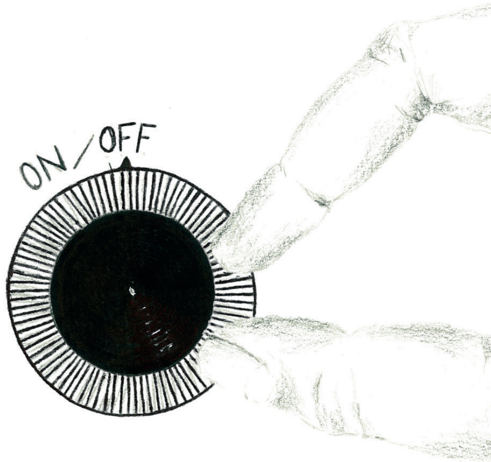
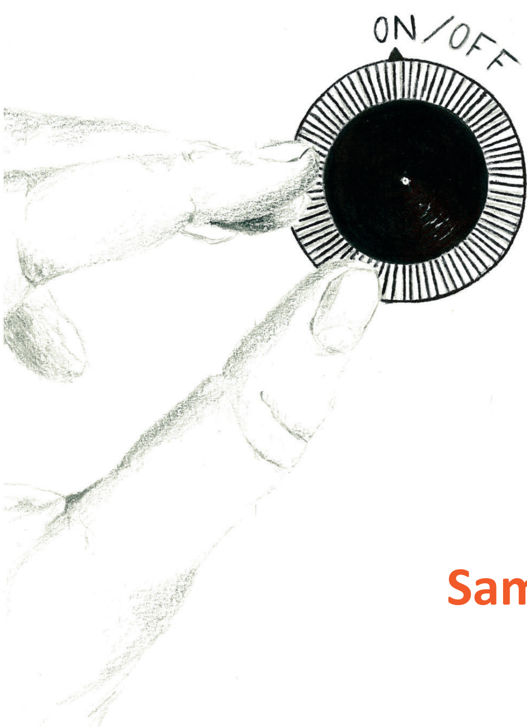
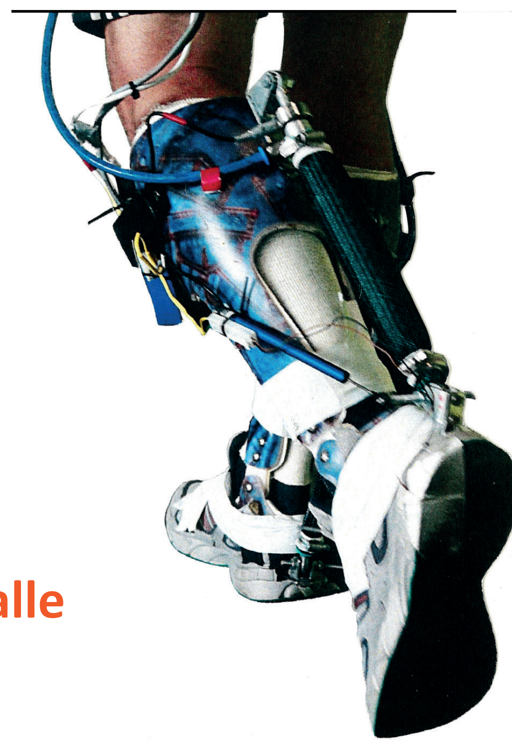


Experimental optimization of an ankle-foot exoskeleton to reduce the metabolic cost of walking for practical applications in healthy and impaired subjects





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UNIVERSITEIT GENT

Faculty of Medicine and Health Sciences
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Experimental optimization of an ankle-foot exoskeleton to reduce the metabolic cost of walking for practical applications in healthy and impaired subjects

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"Walking is for enjoying from one day to another …." (Winter, 1991)

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waarbij ik al mijn hele leven met alles terecht kan en waar ik ook nog eens twee fantastische kinderen en een even fantastische man kan vinden. Ook oma en Peter ben ik dankbaar voor de vele mooie momenten en de steun, mijn peter Dirk voor vele leuke momenten en de vele andere familieleden die telkens opnieuw voor aangename moment zorgen.

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SAMENVATTING

INTRODUCTIE

Mensen hebben zich nooit kunnen neerleggen bij de beperkingen van hun eigen lichaam. Een van de mogelijke manieren om de menselijke prestaties te verbeteren of de prestaties van mensen met een beperking te herstellen, is door het gebruik van een exoskelet. Deze exoskeletten kunnen rond de eigen benen bevestigd worden en assisteren de bewegingen van de gebruiker. In tegenstelling tot apparaten op wielen hebben ze het voordeel dat ze de mogelijkheden van de gebruiker maximaal benutten en ook binnen of op moeilijk terrein gebruikt kunnen worden. Het metabole energieverbruik, of de metabole kost van wandelen, is een belangrijk aandachtspunt in exoskeletonderzoek. Door de complexiteit van het wandelpatroon moet de assistentie van een exoskelet optimaal zijn om de metabole kost van wandelen te verlagen. Voorgaande exoskelet studies slaagden hier niet in doordat er onvoldoende aandacht besteed werd aan de interactie tussen het exoskelet en de gebruiker. Dit leidde tot de ontwikkeling van kleinere exoskeletten die minder zwaar waren en waarbij de assistentie van het exoskelet gemakkelijker te controleren was. Door het belang van de afstoot tijdens wandelen en de verminderde afstoot die bij verschillende pathologieën tot uiting komt, werd potentieel gezien in exoskeletten die de afstoot tijdens wandelen ondersteunen om de metabole kost van wandelen te verlagen en om wandelen te assisteren in populaties met verminderde wandelcapaciteiten.

DOELSTELLING EN RESULTATEN

Deze thesis is gefocust op het optimaliseren van de interactie tussen het exoskelet en de gebruiker om het energieverbruik van wandelen te verlagen en exoskeletten te kunnen gebruiken voor praktische toepassingen bij gezonde mensen en mensen met een beperking. Voor deze thesis werd een assisterend enkel exoskelet (Wearable Assistive Lower Leg eXoskeleton, WALL-X) ontwikkeld en geoptimaliseerd. WALL-X bestaat uit een enkel orthese met een pneumatische spier die het strekken van de enkel ondersteunt tijdens de afstoot in wandelen. Door de externe hardware en energiebron kan WALL-X enkel op een loopband gebruikt worden. We ontwikkelden een controle algoritme dat toeliet om de activering van het exoskelet en het gemiddelde mechanisch vermogen ter hoogte van de enkel te manipuleren. Op die manier kon het exoskelet assisteren van vroeg in de steunfase tot laat in de steunfase en kon het meer of minder enkel vermogen leveren tijdens wandelen. Om de interactie tussen de gebruiker en het exoskelet te optimaliseren hebben we eerst gefocust op het optimaliseren van de adaptatie aan wandelen met een exoskelet en het optimaliseren van de assistentie van het exoskelet. Hiervoor gebruikten we een systematische aanpak waarbij de metabole reductie geoptimaliseerd werd door het onderzoeken van de menselijke reacties op veranderingen in de assistentie van het exoskelet.

In een eerste experiment werd aangetoond dat het ongeveer 20 min duurt alvorens proefpersonen metabool aangepast zijn aan wandelen met WALL-X. Hoewel de proefpersonen kinematisch gezien aangepast waren na enkele minuten, verminderde de spieractiviteit nog verder tijdens de adaptatie zodat de grootste reductie in de metabole kost na ongeveer 20 min gevonden werd. In een tweede experiment werd de assistentie van het exoskelet geoptimaliseerd door de start van de activatie en de hoeveelheid mechanisch vermogen van het exoskelet te optimaliseren. Dit resulteerde in een reductie in de metabole kost van 21% in vergelijking met niet-aangedreven exoskelet wandelen (dit is wandelen met het exoskelet maar zonder de assistentie van de pneumatische spieren). Door de toename in energieverbruik door het dragen van het niet-aangedreven exoskelet bedroeg de reductie in de metabole kost in vergelijking met gewoon wandelen 12%. Ter voorbereiding van een experiment bij maximale inspanning, werd er ook een experiment uitgevoerd op een helling. Door het optimaliseren van de start van de activatie tijdens bergop wandelen was de metabole kost van wandelen met het exoskelet 12% lager in vergelijking met niet-aangedreven exoskelet wandelen. Deze drie experimenten werden uitgevoerd om de assistentie van het exoskelet te verbeteren en de metabole kost van wandelen met het exoskelet te verlagen. Deze verbeteringen lieten ons toe om het exoskelet te gebruiken voor praktische toepassingen bij gezonde mensen en bij ouderen.

Als eerste werd een experiment uitgevoerd waarbij proefpersonen wandelden op een steile helling waarbij ze alsmaar meer gewicht moesten dragen tot ze uitgeput waren. We toonden aan dat het exoskelet op die manier de maximale wandelprestatie kan verbeteren aangezien proefpersonen deze taak langer konden volhouden en meer gewicht konden dragen wanneer het exoskelet hen assisteerde. Er werd ook een experiment gedaan waarbij gezonde ouderen (65+) met het exoskelet wandelden na een korte adaptatieperiode. Er werd aangetoond dat ons exoskelet de metabole kost met 12% kan verlagen bij gezonde ouderen in vergelijking met niet-geassisteerd wandelen met het exoskelet. Ten slotte werd ook een piloottest gedaan om aan te tonen dat patiënten met Chronisch Obstructief Longlijden (COPD) met ons exoskelet kunnen wandelen en er ook voordeel uit kunnen halen.

CONCLUSIE

Met deze thesis toonden we aan hoe het optimaliseren van de adaptatie aan het exoskelet en de assistentie van het exoskelet kan leiden tot significante reducties in de metabole kost van wandelen. Onze bevindingen, in combinatie met bevindingen van andere onderzoeksgroepen, kunnen er voor zorgen dat er in de toekomst ook autonome exoskeletten ontwikkeld worden die het energieverbruik met 20% kunnen doen dalen. Dit moet toelaten om exoskeletten te gebruiken voor praktische toepassingen, ook buiten en in levensechte situaties. Zo toonden we aan dat exoskeletten kunnen gebruikt worden om de maximale prestatie te verbeteren bij gezonde personen of om het metabole

energieverbruik te verlagen bij ouderen en zelfs bij patiënten met COPD. Toekomstig onderzoek moet verder bouwen op onze resultaten proberen om exoskeletten nog verder te verbeteren en het gebruik van deze exoskeletten bij personen die weinig inspanning kunnen verdragen verder onderzoeken. Op termijn moeten exoskeletten in staat zijn om de levenskwaliteit van bepaalde doelgroepen te verbeteren en we hopen dat onze bevindingen daar kunnen aan bijdragen.

SUMMARY

INTRODUCTION

Humans have never been able to accept the limitations of their bodies. One way to improve human performance in healthy subjects or restore performance for impaired subjects is the use of exoskeletons to assist locomotion. These mechanical devices can be worn by the user, are closely attached to the body and work in concert with the movements of the user. Exoskeletons have some advantages over wheeled devices as they still depend on the abilities of the user and could be worn indoors or on rough terrain. The metabolic energy use or the metabolic cost of exoskeleton walking is an important concern in exoskeleton research. Due to the complexity of the already highly efficient walking pattern, exoskeleton assistance needs to be optimal in order to reduce the metabolic cost of walking. Earlier exoskeletons failed in reducing the metabolic cost of walking because of insufficient attention on the human-exoskeleton interaction, leading to the development of smaller exoskeletons that added less weight to the user and allowed easier control of exoskeleton assistance. Due to the importance of the ankle push-off during walking, and the reduced ankle-push-off that is seen in several pathologies, exoskeletons that assist the push-off during walking were believed to be able to reduce the metabolic cost of walking and have potential for gait assistance in impaired populations.

AIM AND RESULTS

This thesis focussed on optimizing the human-exoskeleton interaction to reduce the metabolic cost of walking in order to use exoskeletons for practical applications in healthy and impaired subjects.

A Wearable Assistive Lower Leg eXoskeleton (WALL-X) was developed and improved in this thesis. WALL-X consists of an ankle-foot orthosis powered with pneumatic muscles that assist plantar flexion during the push-off in walking. Due to the off-board hardware and power source, WALL-X can only be used on treadmill. We developed a control algorithm that allowed to control the actuation timing and the average exoskeleton ankle joint mechanical power so that exoskeleton assistance could be manipulated to assist from early in the stance phase to late in the stance phase and to deliver low or high amounts of ankle power during walking.

To optimize the human-exoskeleton interaction, we first focussed on optimizing the adaptation to exoskeleton walking and optimizing the exoskeleton assistance. We used a systematic approach where metabolic reductions were optimized by exploring human responses to changes in different exoskeleton assistance parameters. In a first experiment, it was shown that it takes app. 20 min before subjects are metabolically adapted to walking with WALL-X. Although subjects seemed kinematically adapted after a few minutes of walking with the exoskeleton, muscle activity was further reduced during the adaptation, resulting in the largest reduction in the metabolic cost after app. 20 min. In a second experiment, exoskeleton assistance was optimized by optimizing the actuation onset timing and the average amount of exoskeleton ankle joint mechanical power, resulting in a reduction in the metabolic cost of 21% compared to unpowered walking (walking with the exoskeleton without the assistance of the pneumatic muscles). Due to the penalty of wearing the exoskeleton, the reduction in the metabolic cost for exoskeleton walking compared to normal walking without an exoskeleton was 12%. In preparation for a maximal performance experiment, also an uphill walking experiment was done. By optimizing actuation onset timing of the pneumatic muscles during uphill walking, the metabolic cost of powered exoskeleton walking was 12% lower compared to unpowered exoskeleton walking. These three experiments were done to improve exoskeleton assistance and reduce the metabolic cost of exoskeleton walking. These improvements allowed us to use the exoskeleton for practical applications in healthy subjects and in elderly.

First, an experiment was performed in which subjects walked on a steep inclination while gradually carrying more weight until exhaustion. It was shown that exoskeletons can improve the maximal walking performance as subjects could maintain the maximal walking exercise test longer and carry a higher weight during exoskeleton assistance. An experiment in which healthy elderly (65+) walked with the exoskeleton after a short adaptation period was also done and showed that our exoskeleton can reduce the metabolic cost of walking compared to unpowered exoskeleton walking with 12% in healthy elderly. At last, a pilot test was done to show that patients with Chronic Obstructive Pulmonary Disease (COPD) can also walk with and benefit from an exoskeleton.

CONCLUSION

In this thesis we showed how optimizing exoskeleton adaptation and exoskeleton assistance could lead to significant reductions in the metabolic cost of walking. Our findings, in combinations with recent findings of other research groups, should allow to develop exoskeletons that are fully autonomous and reduce the metabolic cost of walking with 20% compared to normal walking in the future. This should allow to use exoskeletons for practical applications, also outdoors and in real life situations. We showed how exoskeletons could be used to increase maximal walking performance in healthy subjects or reduce the metabolic cost of walking in elderly and even in COPD patients. Future research should built on our results to further optimize exoskeleton assistance and to study the use of exoskeletons in populations with reduced exercise tolerance. Somewhere in the near future, exoskeletons will allow to improve quality of life of certain populations and we hope that this thesis contributed to this goal.

INTRODUCTION

INTRODUCTION

1. Introduction on exoskeletons

In humans and other vertebrate animals, locomotion patterns are optimized to minimize energy requirements (Alexander, 1989). In contrast to other animals, humans have never been able to accept the limitations of their bodies and searched for alternatives to cover longer distances with less effort. Humans always wanted to *“… work it harder, make it better, do it faster, makes us stronger …”*, just like in the song ‘Harder, better, faster stronger’ by the electronic band Daft Punk.

This lead to the development of the first bicycle in 1817 (Lessing, 2003), the first patent on a car in 1886 (Benz & Co, 1886) and the first plane in 1903 (Kelly, 1989). Around that time, Nicholas Yagn patented his ‘apparatus for facilitating walking, running and jumping’ (Yagn, 1890) (Fig. 1). The objective, as stated in the original patent application, was to increase the efficiency of walking, running and jumping and further decrease the fatigue inherent to the act of walking, running, and jumping. Such an ‘apparatus’, that acts in parallel with the human body to assist human locomotion, is now called an exoskeleton as it could be considered an additional skeleton that is applied to the body. Exoskeletons are defined as mechanical devices that are essentially anthropomorphic in nature, are 'worn' by an operator, fit closely to the body, and work in concert with the operator's movements (Dollar and Herr, 2008).

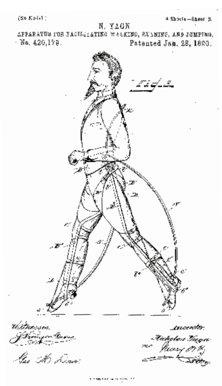

Fig. 1
Apparatus for facilitating walking, running and jumping, patented in 1890 by Nicholas Yagn. From Yagn (Yagn, 1890).

The ultimate goal of such an exoskeleton is to improve human performance for healthy subjects or restore performance for impaired subjects. Despite the functional goal of the exoskeleton (which could be to increase speed, increase walking duration, reduce the effort, improve walking symmetry, etc.), reducing the metabolic cost of walking with an exoskeleton should be a major concern in exoskeleton development (Ferris et al., 2007; Guizzo and Goldstein, 2005). Reducing the metabolic cost can be the functional goal of the device, in example for subjects for which normal walking is already a heavy effort due to a disease or ageing, but will also increase the usability and the number of possible applications, both for healthy and impaired subjects. Reducing the metabolic cost of walking with the exoskeleton will also have a positive effect on other functional goals like an increase in walking speed, walking duration, etc.

Back in 1890, Nicholas Yagn had a detailed plan on both the design and the functioning of his exoskeleton (Yagn, 1890). While bikes, cars and planes have evolved in the last decade and became a part of our daily life, the use of an exoskeleton to improve locomotion is rather inexistent. The most surprising is the fact that it seems very difficult to successfully interact with the human body during locomotion. Given the huge technical evolution in the last decade, technology does not seem to be the limitation. Humans can walk without the need to think how to walk but walking is still a complex motor task which is hard to mimic if we look to the rigid walking pattern of most humanoid robots. To improve this complex motor task with an exoskeleton, the interaction between the human mechanism and the mechanical assistance of an exoskeleton needs to be optimized. However, human-exoskeleton interaction received only little attention in the last decade.

Therefore, this thesis focusses on the optimization of the human-exoskeleton interaction during walking from a biomechanical and physiological point of view in order to reduce the metabolic cost of walking with exoskeletons and improve the understanding of the human-exoskeleton interaction in general so that exoskeletons can be used as a tool to assist walking in the (near) future. The approach where scientists from different backgrounds work on exoskeletons, each approaching the challenges from their expertise and sharing their knowledge, thereby improving general understanding of assisted movement, is essential to advance exoskeleton research (Ferris et al., 2007) as the main conclusion since the patent of Nicholas Yagn is probably that it is not easy to improve human walking.

This thesis focusses on optimizing the human-exoskeleton interaction in order to reduce the metabolic cost of exoskeleton walking.

2. Focus on walking

"Walking is for moving from one place to another ..." (Winter, 1991)

Walking is the most frequent way of locomotion in humans and the second most important way of transport. In the western world we make 25 to 30% of our displacements on foot (Mitchell, 2006). The ability to walk autonomously has a major influence on mobility, which is related to quality of life and subjective well-being (Ravulaparthy et al., 2013). This becomes obvious when one loses the ability to walk because of an accident, disease or age. Almost everybody walks on a daily basis to move from one place to another. One could say that walking is connected to the 'circle of life'. Children start walking around the age of 1 and show characteristics of adult gait within one year after the initiation of independent gait (Burnett and Johnson, 1971). The walking pattern further develops and during adulthood people walk rather subconscious, taking almost 10000 steps a day (Bohannon, 2007), thereby making walking an important part of daily life. With age the number of steps reduces again (Bohannon, 2007) and the walking pattern deteriorates (Winter et al., 1990) until we again lose the ability to walk because of diseases, age or death.

There can be several reasons why walking is sometimes not the preferred way of locomotion, i.e. because the human system, the task or the terrain does not allow to walk easily. An exoskeleton could be a tool to overcome these constraints and improve the walking abilities in a large number of situations, both for healthy or impaired people. It also has the advantage that a person still uses their own capabilities, which is less the case in wheeled devices. Because of the frequency of walking in daily life and the importance of the ability to walk independently, this thesis focusses on exoskeleton assistance for walking. To successfully assist walking with an exoskeleton, it is necessary to understand the general walking pattern. Therefore, the next pages will focus on the mechanics and energetics of walking, with specific attention on aspects that are essential for walking assistance with exoskeletons.

This thesis focusses on walking assistance with exoskeletons, for which a good understanding of the walking mechanics and energetics are essential.

2.1. Describing the walking pattern

"Walking is very important for meeting the world …" (Winter, 1991)

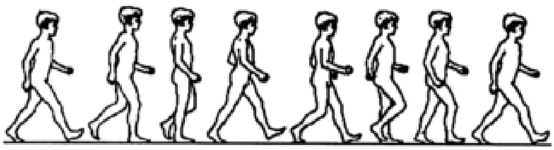

Fig. 2
Representation of movements in the sagittal plane for a walking stride from heel contact of the right leg to the next heel contact of the right leg. From Novacheck (Novacheck, 1998).

Walking is a phenomenon of the most extraordinary complexity. The body is supported by alternating the left and the right leg and body weight is transferred from one leg to the other. Within each leg, a stance phase (from initial contact until toe-off) is followed by a swing phase (from toe-off until the next initial contact) and at every instant there is at least one foot in contact with the ground in alternating single stance and double stance phases. During normal walking the left and the right leg behave symmetrical with a 50% phase shift, indicating that the stride of the left leg starts when the right leg is at 50% of its stride and vice versa. The stance phase lasts around 60% of the stride time and the swing phase around 40% of the stride time, resulting in a double support phase with a duration of around 10% of the stride time (Novacheck, 1998). The stance phase could be further divided into three main functional phases. The first 15% of the stride is called the weight acceptance phase, midstance is then situated between 15% and 40% of the stride and the end of the stance phase is called the push-off, between 40% and 60% of the stride, during which the lower limb pushes away from the ground (Winter, 1991). The swing phase could further be divided into two functional phases. Mid swing is the midpoint in time between toe-off and initial contact, and divides the swing phase into early swing or lift-off and late swing or reach (Winter, 1991).

The focus of the description of walking will be on movements of the legs in the sagittal plane during the stance phase because these are the major movements that drive walking. During walking, movements in the ankle joint occur mainly in the sagittal plane. The ankle joint is in almost neutral position at heel contact and shows rapid plantar flexion during foot placement, followed by increasing dorsiflexion during midstance (Fig. 3). During the push-off, maximal plantar flexion occurs when the leg pushes away from the ground and the ankle joint returns to a neutral position during the swing phase to prevent the foot to touch the ground and to prepare the next heel contact (Winter, 1983). There is also a small shift in the rearfoot from eversion in the majority of the stance phase through inversion during the push-off and from abduction in the majority of the stance phase to adduction during the push-off (Moseley et al., 1996; Rattanaprasert et al., 1999). In the knee joint, the major movements also occur in the sagittal plane. The knee joint is in almost extended position at heel contact, followed by little flexion during the weight acceptance phase and knee extension during midstance. Flexion arises during the push-off and maximal flexion occurs during the swing phase to prevent the foot to touch the ground (Fig. 2). There is limited varus or adduction during the stance phase and a shift from limited internal to external rotation during the stance phase (Kadaba et al., 1990; Yu et al., 1997). The hip joint is in maximal flexion around heel contact as the foot is in front of the hip and moves towards maximal extension during push-off as the foot is behind the hip during this phase. During the swing phase the hip angle again goes from extension to flexion (Fig. 2). Apart from these sagittal movements there is also pelvic rotation (Kerrigan et al., 2001; Saunders et al., 1953), pelvic tilt (Gard and Childress, 1997) and pelvis lateral displacement (Donelan et al., 2004) during normal walking.

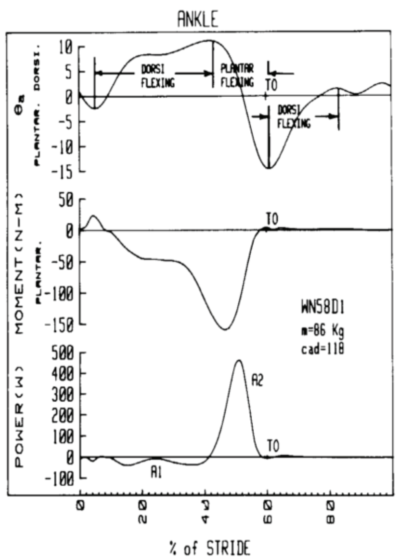

Fig. 3

Joint angle, joint moment (which is the same as the joint torque) and joint power of the ankle joint in the sagittal plane from a representative subject of 86kg during natural walking on an over ground walkway. A1 is representing the negative work during midstance, where an extension torque is present during ankle flexion. A2 represents the positive work during the push-off, where an extension torque is present during ankle extension. TO refers to toe-off. From Winter (Winter, 1983).

During walking, forces are applied between the feet and the ground to maintain an upright position and go forward. The individual leg horizontal ground reaction force pattern for walking acts decelerating in the first half of the stance phase, due to the dorsal position of the body centre of mass (bCOM) relative to the foot. It crosses zero during midstance and acts accelerating in the second half of the stance phase due to the ventral position of the bCOM relative to the foot (Nilsson and Thorstensson, 1989). The vertical ground reaction force shows a typical ‘double hump’ pattern (Nilsson and Thorstensson, 1989), with the first hump being the result of the weight acceptance phase, where the downwards velocity of the body is decreased. The second hump is the result of the push off, where the bCOM is accelerated upwards. The medio-lateral forces are much smaller compared to the vertical and horizontal forces (Nilsson and Thorstensson, 1989).

The torque is the product of a force acting at a distance about an axis of rotation and causes an angular acceleration about that axis (Winter, 1991). During walking, the joint torques are the net result of the muscles and other structures such as tendons and joint frictions that are acting to alter the angular position of the joints, also referred to as internal joint moments. These joint torques thus underestimate the actual forces that are acting on a joint due to co-contraction and active stabilisation.

Joint powers are calculated as the product of the net joint torque and the joint angular velocity and are used to quantify the rate of generating or absorbing energy by the muscles or the work performed per unit time (Winter, 1991). Work is than defined as the numerical integration of power over time, or the area under the power curve. Joint torques and joint mechanical powers help to interpret the walking mechanics. During the stance phase, the ground reaction force vector passes close to the ankle, knee and hip joint and minimizes the joint torques (Vaughan, 1996).

In the ankle, more than 90% of the work is performed in the sagittal plane (Eng and Winter, 1995). In the knee, there is a considerable amount of work performed in the frontal plane (10%) but the majority of the work is still performed in the sagittal plane. In the hip, the largest amount of work is also performed in the sagittal plane but also a large amount of work (25%) is performed in the frontal plane (Eng and Winter, 1995).

Overall, the largest sagittal joint torques are found in the ankle joint during walking, with a plantar flexion torque during almost the entire stance phase (Fig. 3). Because of the high plantar flexion velocity during the push-off and the high joint torques in the ankle, peak power generation (Fig. 3) is highest in the ankle. The area under the positive peaks of the power curve (A2 in Fig 3), the positive work, is also largest in the ankle joint when compared to the knee or hip joint (Fig. 4). The area under the negative peaks of the power curve (A1 in Fig 3) is rather small. During normal walking, more than 50% of the total positive work is done by the ankle (Farris and Sawicki, 2012a; Novacheck, 1998). Phases of positive and negative work are also interpreted as phases of mechanical energy generation and energy absorption. The plantar flexor muscles, that absorb mechanical energy during the weight acceptance phase and the midstance phase, show a dominant energy generation burst during the push-off (Winter, 1983). This is the major energy that propels the body forward during walking as the plantar flexion power is translated into kinetic and potential energy of the bCOM (Winter, 1983).

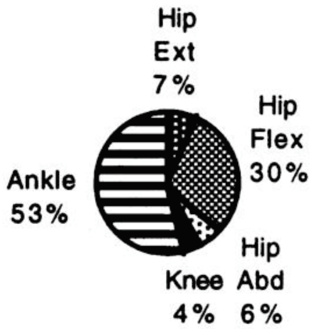

Fig. 4
The distribution of positive work (area underneath the power curve positive peaks) in the leg during walking at 1.2 m·s^{-1}. Positive work is largest in the ankle joint (53%), followed by positive work during hip flexion (30%). From Novacheck (Novacheck, 1998).

While the focus of this thesis will be on lower limb movements in the plane of progression, walking is not a one dimensional movement. Early work by Saunders et al. (Saunders et al., 1953) suggested 6 major determinants of human walking, such as pelvic rotation, pelvic tilt and pelvic lateral displacement, which consists of movements in the sagittal, frontal and transversal plane. Trunk movements also play an important role in walking. Inverse dynamics indicated a strong influence of joint moments in the lower limb on trunk angular accelerations (Nott et al., 2010). Also, the trunk has a major role in reducing the translation and rotational displacements and accelerations of the head relative to the pelvis and shows variations in trunk angle between -1° and +1° during walking (Winter, 1991). To control balance in the frontal plane during walking, the trunk and pelvis play an important role on 2 levels. First to balance the arms, head and trunk around the supporting hip and second to balance the total body centre of mass around the supporting foot (MacKinnon and Winter, 1993). These pelvic and trunk movements are especially important for balance control during walking (Winter, 1995) and work in synergy with movements in the support foot (MacKinnon and Winter, 1993).

A description of the walking pattern shows that the ankle plays an important role during walking, being the largest source of positive joint work, performed during the push-off.

2.2. Muscles: the walking motors

" We see others walking [...] and we want to do it too." (Winter, 1991)

The description of the walking pattern shows the importance of the ankle push-off during walking. However, the muscles-tendon complexes are responsible for the power that is necessary for locomotion and surface electromyography (EMG) allows to measure what happens on a muscular level during walking in the most important leg muscles (Fig. 5) (Whittle, 2007; Winter, 1991). Muscle activity patterns during walking show some differences between studies, which could be due to electrode placement, differences in the experimental protocol and/or differences in the analysis procedure. A general pattern for the most important leg muscles is given in Fig. 6.

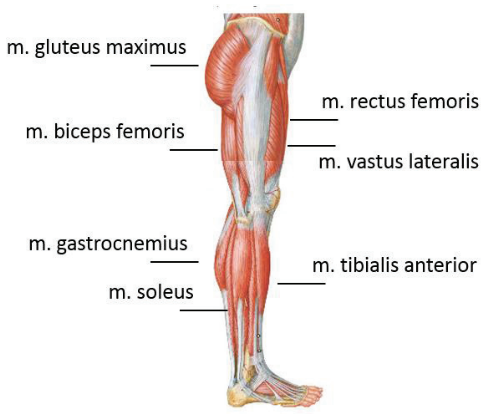

Fig. 5
Anatomical location of the most important leg muscles for walking that causes movements around the ankle, knee and hip joint. Surface EMG allows to measure the muscular activity during locomotion. Adapted from Netter (Netter, 2003).

The m. gluteus maximus extends the hip and is mainly active during the weight acceptance phase to control hip flexion (Fig. 6). A smaller burst arises in the swing phase to decelerate the forward swing of the thigh. The m. rectus femoris, which is a bi-articular muscle and part of the quadriceps muscle group together with the m. vastus medialis, the m. vastus intermedius and the m. vastus lateralis, also shows major activity during the weight acceptance as a knee extensor to control knee flexion and for knee extension during midstance. A smaller peak is found in early swing to assist hip flexion to swing the leg forward and for knee extension to decelerate the backwards swinging leg and foot. The hamstrings, which are also bi-articular and formed by the m. semimembranosus, the m. semitendinosus and the m. biceps femoris, serve as knee flexors at the end of swing to decelerate the leg and foot and as hip extensors around heel contact. The soleus muscle and the gastrocnemius muscle, a bi-articular muscle responsible for ankle plantar flexion and knee flexion, show peak activity during the push-off to initiate rapid plantar flexion. The m. tibialis anterior, part of the anterior tibial muscle group, which is responsible for dorsiflexion, is highly active after heel contact to control plantar flexion and in the beginning of swing to initiate toe clearance during swing.

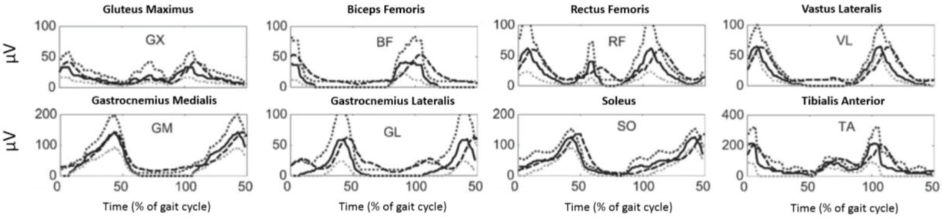

Fig. 6
Muscle activity patterns during normal walking for 9 and 11 subjects (depending on which muscle) of Hof et al. (Hof et al., 2005) as solid black lines, combined with 18 subjects by Winter (Winter, 1991) as dashed lines for the most important leg muscles during walking. Data are shown for one and a half stride to emphasize muscle activity around heel contact. See Fig. 5 for anatomical muscle locations. Dotted lines are the upper and lower limits for 'standard EMG-profiles' during walking. Most activity is seen in the beginning of the stance phase (around heel contact) and around 50% (when the push-off occurs). Adapted from Hof et al. (Hof et al., 2005).

Overall, the muscular activity during swing is low compared to the muscular activity during the stance phase. The plantar flexor muscles have the most important role in forward progression and are the most important sources of positive muscle work during walking (Gottschall and Kram, 2003; Meinders et al., 1998; Neptune et al., 2001; Winter, 1983). These plantar flexor muscles initiate movements between the foot and the lower leg, mainly by rotation at the talocrural joint (between tibia, fibula and talus) and to a smaller degree at the talocalcaneal joint (between talus and calcaneus) (Arndt et al., 2004). Still, the ankle joint cannot be described as a simple hinge joint as the ankle joint axis alters during the stance phase in walking (Arndt et al., 2004).
The plantar flexor muscles are often analysed as a single unit, although the soleus and gastrocnemius muscle have different functions in specific instances of the gait cycle (Neptune et al., 2001). As the gastrocnemius is a bi-articular muscle covering the ankle and knee joint, it can generate knee flexion moments and ankle plantarflexion moments while the soleus can only generate plantarflexion moments. Some studies suggested that the soleus and gastrocnemius muscles can generate opposite acceleration at the hip and similar knee extension by dynamic coupling (Neptune et al., 2004a) while others found that the soleus muscle causes ankle plantarflexion and knee extension, while the gastrocnemius muscle causes ankle dorsiflexion and knee flexion (Lenhart et al., 2014; Stewart et al., 2007). The soleus and gastrocnemius muscles also play an important role in support and forward progression of the trunk during walking (Zajac et al., 2003).

During walking the triceps surae muscle group, which consists of the gastrocnemius and the soleus muscle and share the Achilles tendon, stores and recoils energy. The long series elastic elements allows the contractile elements to act isometrically during the stance phase and allows the triceps surae to shorten at high velocities during the push-off. This improves efficiency when compared with more proximal muscles (Ishikawa et al., 2005; Lichtwark et al., 2007; Roberts, 2002; Sawicki et al., 2009). Several studies indicate that around 50% of the positive mechanical work by the plantar flexors could be attributed to elastic recoil of the Achilles tendon and 50% could be attributed to active shortening of the contractile elements (Farris and Sawicki, 2012b; Fukunaga et al., 2002; Ishikawa et al., 2005; Lichtwark and Wilson, 2006).

Experimental analyses showed that positive muscle work is metabolically more expensive than isometric muscle work and that isometric muscle work is more expensive than negative muscle work (Rall, 1985). Although isometric and negative muscle work require some metabolic energy cost (Ryschon et al., 1997), positive muscle work during walking causes the majority of the metabolic energy cost of walking (Umberger and Martin, 2007). However, due to the combination of isometric, positive and negative muscle work and co-contractions between muscles, it is hard to calculate total energy expenditure based on muscle work. Also, muscular efficiency for positive and negative mechanical work could be estimated around respectively 25% and -120% (Margaria, 1968) but total mechanical work (Sasaki et al., 2009) or total joint work (Umberger and Martin, 2007) calculations underestimate muscle tendon work. Due to energy storage and return and bi-articular muscles, the relation between joint work and muscle work during locomotion remains unclear (Ferris et al., 2007).To fully understand the cost of locomotion one should be able to do an analysis of all external forces (weight, ground reaction forces, bony forces, muscles forces, ligament forces, skin forces) acting on a segment and a complete analysis of the energetics of force and work production by the muscles. However, practical and ethical limitations do not allow to do so.

Muscular activity patterns emphasizes the importance of the plantar flexor muscles, being the largest source of muscle work, mainly during the push-off.

2.3. About simple models and mechanical work

"… walking does not come automatically…" (Winter, 1991)

Apart from describing walking, several models have been made to represent walking in order to get a better understanding of the mechanics and energetics. Most of them use an inverted pendulum (Fig. 7) that conserves mechanical energy. Walking is then modelled with straight massless legs with the bCOM placed at the hip and moving forward in a series of circular arcs with a radius equal to the leg length (Alexander, 1976; Cavagna and Margaria, 1966; Cavagna et al., 1963). With also the swing leg largely behaving as a passive pendulum (Mochon and McMahon, 1980), walking theoretically does not require mechanical work as there is no dissipative load external to the body and no net work against gravity is performed (Donelan et al., 2002a). A bipedal spring-mass model, which describes the legs as two massless, linear springs, still uses an inverted pendulum but also shows how the walking energetics can be combined with a double support phase and a realistic ground reaction force pattern without necessary energy input (Geyer et al., 2006).

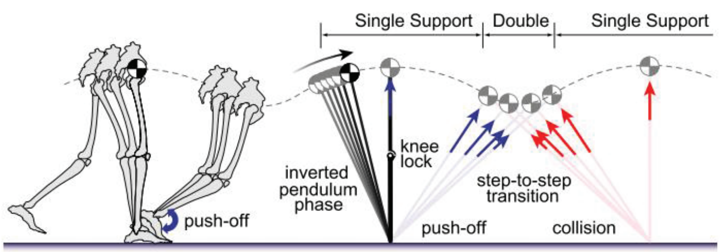

Fig. 7
Representation of the inverted pendulum mechanism during the single support phase, resulting from knee lock, which only requires minimal force or work. During the double support phase, the bCOM velocity must be redirected in the step-to-step transition by positive work performed by the trailing leg, the push-off work, and negative work performed by the leading leg, the collision work. From Kuo (Kuo, 2007).

It should be kept in mind that external work measurements are not directly related to the energetic cost of muscles as these neglect mechanical energy changes of the body segments, do not take the possible simultaneous positive and negative muscle work in consideration and overlook elastic energy storage and release (Kuo and Donelan, 2009; Neptune et al., 2004b). Still, it can help to improve our understanding of walking. Experimental analysis of the mechanical work during walking merely focusses on external work, referring to the work associated with mechanical energy changes of the bCOM. Early observations of human walking suggested that bCOM vertical movements should be reduced (Perry,

1992) due to the mechanical work that is associated with abrupt changes in the direction of the bCOM motion. However, the inverted pendulum mechanism that arises during walking, where kinetic and gravitational potential energy of the bCOM are in opposite phase (Fig. 8), allows energy transfer between kinetic and gravitational potential energy (Farley and Ferris, 1998) and explains how bCOM movements can still be energetically beneficial. Analyses of the external mechanical work gave insight into the recovery of mechanical energy during the single stance phase in walking, which can be up to 70% at intermediate walking speeds (Cavagna and Heglund, 1977). However, there is still a considerable mechanical energy cost during the double support phase when the bCOM velocity needs to be redirected (Alexander, 1995).

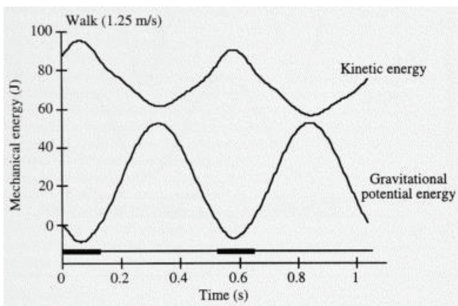

Fig. 8

Kinetic and gravitational potential energy fluctuations for a stride during walking at 1.25 m·s^{-1}, which indicates a pendulum-like energy interchange during the single stance phase in walking. From Farley & Ferris (Farley and Ferris, 1998).

Dynamical walking approaches, referring to “systems in which the passive dynamics of the limbs dominate the motion, with minimal actuation applied to sustain periodic behaviour” (Kuo, 2007), help to understand how mechanical work must still be performed during walking. While the single stance phase is energetically efficient due to the inverted pendulum mechanism, the transition from one step to another, in which the bCOM velocity needs to be redirected, is costly and is called the step-to-step transition (Donelan et al., 2002a; Kuo, 2002). Research with forward dynamics modelling (Neptune et al., 2004b) suggested that maintaining the pendulum-like motion is also costly but this hypothesis is controversial (Kuo and Donelan, 2009).

Even in simple models, walking on level ground requires mechanical energy input to overcome the energy loss during the step-to-step transition in order to redirect the bCOM, which is called the collision cost. Passive dynamic walkers (Collins, 2001; McGeer, 1990), where a bilateral simple mechanism walks down a slope without external actuation or control, shows that gravity can compensate for this collision cost. Kuo (Kuo, 2002) showed that in simple models the collision cost could also be compensated for by an impulsive push along the trailing leg, immediately before heel strike. This corresponds to results in passive dynamic robots (Collins et al., 2005) and to the biological ankle behaviour in walking, in which the collision cost is compensated for with toe-off impulses just before and during the double stance phase (Kuo and Donelan, 2010). While collision is an instantaneous event in simple models, collision in human walking is characterized by performing negative work over a collision phase, which starts around heel strike and lasts beyond double limb support (Kuo and Donelan, 2010)(Fig. 9). The collision phase results in a metabolic energy cost because of the negative work done by the leading leg to redirect the bCOM velocity and the positive work done by the trailing leg in the pus-off to restore the energy loss during double support (Donelan et al., 2002b).

Experimental work revealed that the mechanical work that is required to redirect the bCOM during the step-to-step transition is a major determinant of the metabolic cost, rather than the pendular motion itself (Donelan et al., 2002a, 2002b; Kuo et al., 2005). Based on these simplified models of human walking, the net metabolic energy cost of walking could then be explained by the step-to-step transition cost, being responsible for around two third of the metabolic energy cost of walking (Donelan et al., 2002a; Kuo et al., 2005), and the cost of the forced motion of the legs during swing (Doke et al., 2005; Kuo, 2007), accounting for around one third of the net metabolic energy cost of walking. The combination of the energetic cost of step-to-step transitions, which increases with step length and step width (Donelan et al., 2001), and the cost of leg swing, which increases with step frequency, can then explain the preferred speed-step length relationships in humans (Kuo, 2001).

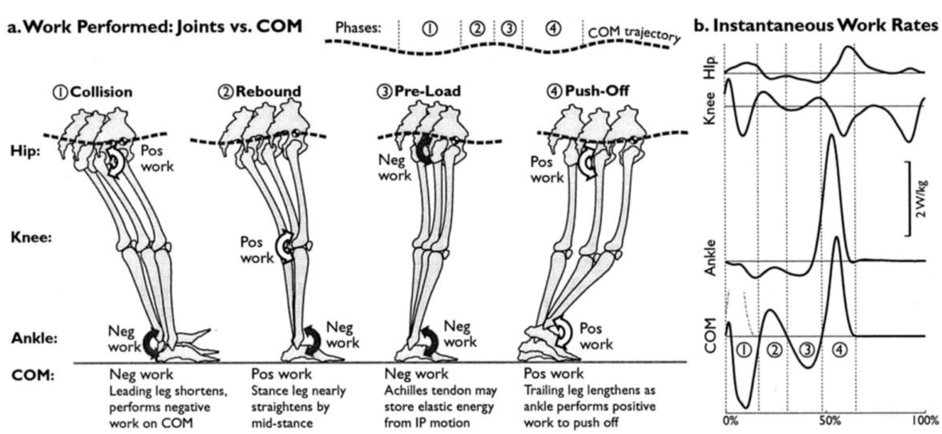

Fig. 9

Representation of the four functional phases during a walking stride based on bCOM power (a): the collision phase, the rebound phase, the pre-load phase and the push-off phase. Area under the bCOM power curve (b) represents the bCOM work for the collision phase, the rebound phase, the preload phase and the push-off phase. Joint powers (or instantaneous work rates) in the ankle, knee and hip help to interpret the bCOM power curve. From Kuo et al. (Kuo et al., 2005).

Joint powers in the ankle, knee and hip could then be related to bCOM work (which is the area under the positive and negative bursts of the bCOM power curve) and could be combined with the findings from simple walking models, resulting into a division of the stance phase in four functional phases during a walking stride (Kuo and Donelan, 2010; Kuo et al., 2005).

Around heel contact, the negative bCOM work in the collision phase is first performed by the ankle and then by the knee although the remainder is suggested to be passively dissipated (Fig. 9). The next phase is called the rebound phase in which positive bCOM work is done by leg straitening and possible (elastic) rebound of the knee, corresponding to muscle activity in the quadriceps muscles (Fig. 6). The last two phases are the pre-load and the push-off phase. In the pre-load phase, negative bCOM work occurs as elastic energy is stored in the Achilles tendon, which will be recycled during the push-off where positive bCOM work is performed by both elastic energy recoil and shortening of the contractile elements in the plantar flexor muscles. The positive work of the push-off in the trailing leg overleaps with the collision phase in the leading leg and compensates for much of the negative work in the collision phase. The push-off phase starts a little before double support and the collision phase extends a little beyond double support (Donelan et al., 2002b). Much of the muscle work during walking, mainly in the ankle, could be seen as reducing the step-to-step transition cost (Donelan et al., 2002a) as Kuo showed that an impulsive push around toe-off can reduce the collision in the leading leg (Kuo, 2002).

Walking compares to an inverted pendulum and mechanical work is mainly necessary during the transition from one step to another, called the step-to-step transition.

2.4. Metabolic energy expenditure

"Walking […] is important for almost everything – like breathing." (Winter, 1991)

As mentioned earlier, mechanical work analyses and muscle work analyses are limited to estimate metabolic energy requirements during walking. At normal walking speeds, walking is an aerobic activity and metabolic energy expenditure, which is an important concern in human locomotion, can be estimated based on O_2 consumption and CO_2 production (Brockway, 1987) during steady state locomotion once oxygen consumption reaches sufficient levels to meet the energy requirements of the muscles. To establish the exact reasons for the metabolic energy expenditure during walking, this energy expenditure is often diminished with the energy cost for standing rest, referred to as the net metabolic energy cost.

The net metabolic energy cost of walking results from the energy cost of performing muscle work. Based on the simple walking models discussed earlier, around two thirds of the net metabolic energy cost is related to the step-to-step transition (Donelan et al., 2002a; Kuo and Donelan, 2010). However, experimental work suggested that the true cost to perform work on the centre of mass during walking is rather around 50% of the net metabolic cost of walking (Grabowski, A. et al., 2005). Simple models also assume rigid legs but experimental work revealed that there is a significant cost of almost 30% of the net metabolic cost of walking associated with muscle force to support body weight (Grabowski, A. et al., 2005). Another important contributor to the metabolic cost of walking is the cost associated with swinging the legs, which is assumed to be between 10 and 30% of the net metabolic cost of walking (Doke et al., 2005; Gottschall and Kram, 2005; Griffin et al., 2003; Neptune et al., 2004b). Recent findings suggest that part of the energy necessary for swinging the legs also results from the ankle push-off (Lipfert et al., 2014), which could explain some studies that reported lower costs associated with leg swing (Gottschall and Kram, 2005). Musculoskeletal models on the other hand suggest that around one third of the metabolic energy cost is related to the step-to-step transitions, one third to the rise and fall of the bCOM during the remainder of the single stance phase and one third to leg swing (Umberger, 2010). Despite the differences between simple walking models, experimental work and musculoskeletal models, the costs associated with mechanical work during the step-to-step transition, swinging the legs and supporting body weight together account for most of the metabolic energy cost of walking. Other smaller contributors are in example active lateral stabilisation (Donelan et al., 2004) and swinging the arms (Collins et al., 2009; Meyns et al., 2013; Ortega et al., 2008).

During normal walking speeds, walking is the most economic gait mode (Hreljac, 1993) and at preferred walking speeds, humans walk at a metabolically optimal step length, step frequency and step width combination (Donelan et al., 2001; Farris and Sawicki, 2012a; Kuo, 2001; Zarrugh et al., 1974). Lower or higher speeds result in a higher metabolic cost of transport, although the speed-energy expenditure curve is nearly flat around the optimum (Cavagna and Kaneko, 1977; Larish et al., 1988; Martin et al., 1992). The metabolic optimum could be attributed to the efficiency of positive muscle fascicle work, which is highest at intermediate speeds (Cavagna and Kaneko, 1977) and to the muscular efficiency to do positive work, depending on the force-length and force-velocity characteristics of skeletal muscles (Farris and Sawicki, 2012a). Mechanical analyses indicate the importance of the ankle plantar flexors during walking and modelling suggests that these consume around 25% of the metabolic energy of walking (Umberger and Rubenson, 2011).

The efficiency of human walking becomes obvious when the cost of transport for human locomotion is compared with the energy requirements of humanoid robots during locomotion, which is more than 10-fold the cost of human locomotion (Collins and Ruina, 2005). However, passive dynamic walking robots that use the passive dynamics of walking combined with ankle actuation are able to walk at a similar energy cost compared to humans, again emphasizing the importance of ankle push-off and passive dynamics for efficient walking (Collins et al., 2005).

The efficiency of walking results in a low metabolic cost, mainly resulting from mechanical work during the step-to-step transition, leg swing and supporting body weight.

2.5. Changes with speed, grade and carried weight

"Walking is for enjoying from one day to another …." (Winter, 1991)

The previous description of walking mainly focused on level walking at normal speeds. In real life, walking is often accompanied with changes in speed, gradient or even weight carrying, which all have an influence on the mechanics and energetics of walking. Regression formulas show that the metabolic energy cost increases drastically with speed, grade and the weight that is carried by the user (Givoni and Goldman, 1971).

Normal walking speeds are situated between 1.2 and 1.5 $m \cdot s^{-1}$ depending on length, sex and age (Bohannon, 1997) with average walking speeds of healthy subjects between 20 and 40 years around 1.4 $m \cdot s^{-1}$ (Bohannon, 1997). Increases in walking speed increases the metabolic energy expenditure (Givoni and Goldman, 1971; Workman and Armstrong, 1986) and also have an effect on spatiotemporal aspects, kinematics, kinetics and EMG (Kirtley et al., 1985; Nilsson and Thorstensson, 1989; Schwartz et al., 2008). Almost every aspect of gait is sensitive to walking speed (Schwartz et al., 2008) and performed work increases in all joints with increasing speed (Farris and Sawicki, 2012a; Neptune et al., 2008). With increasing walking speeds, the negative work in the ankle slightly reduces and the positive work largely increases as both plantar flexor torque and angular velocity increase with increasing speeds (Winter, 1983). While the ankle joint has to perform considerable amounts of work for both lower and higher walking speeds, this indicates that the contractile elements of the plantar flexors have to perform more work during higher walking speeds.

During uphill walking, metabolic energy expenditure increases with grade (Givoni and Goldman, 1971; Kramer, 2010) as the body needs to perform external mechanical work against gravity (Minetti et al., 1993). Mechanical energy exchange is less effective (Gottschall and Kram, 2006) and uphill walking demands more positive mechanical work in the ankle, knee and hip joint (Lay et al., 2006, 2007; McIntosh et al., 2006). While total joint work increases, the relative contribution of ankle joint work reduces (Lay et al., 2006; McIntosh et al., 2006). However, the ankle still needs to perform a considerable amount of work during uphill walking. As the relative contribution of elastic storage and recoil to ankle joint work reduces from 50% to less than 40% during uphill walking, the contractile elements needs to perform a larger portion of the ankle joint work compared to level walking (Sawicki and Ferris, 2009a).

Walking, especially during leisure activities, often includes carrying a backpack. Carrying weight increases the metabolic energy expenditure during walking (Givoni and Goldman, 1971; Kramer, 2010). During weight carrying, the ground reaction forces and the joint torques increase with increasing load and the walking pattern changes (Knapik et al., 1996, 2004). However, the ankle still needs to perform more than 50% of the positive mechanical power during loaded walking (Huang and Kuo, 2014). Adding weight to the legs instead of a backpack also increases the metabolic energy expenditure, with the increase being larger for more distally located weights (Browning et al., 2007).

Changes in speed, grade or weight have an influence on the mechanics and energetics of walking. However, the ankle is still important during walking in varying environments.

2.6. Changes with age and reduced function

"Walking can make the difference whether we ever stand up straight or not" (Winter, 1991)

With increased age or reduced function, several changes arise in the walking mechanics and energetics. Metabolic energy cost of walking increases (Martin et al., 1992), while muscle mass, cardiovascular fitness and physical capabilities reduce with age (Hawkins and Wiswell, 2003; Milanović et al., 2013; Skelton et al., 1994). The reduction in cardiovascular fitness might be due to reductions in maximal stroke volume, reductions in maximal heart rate, changes in body composition, oxygen utilisation in the muscles, etc. (Hawkins and Wiswell, 2003). This also leads to small aerobic reserves during walking, which is related to the reduced walking speed in elderly (Schrack et al., 2010). Due to the degeneration of the physical and cardiovascular capabilities some elderly suffer to perform daily life activities. As previously discussed, human seek to minimize energy expenditure. Although elderly have an increased metabolic cost compared to young subjects at all walking speeds, the curve of the speed-economy curve does not change a lot and elderly still walk around the most economically speeds, given the age-related changes in musculoskeletal and neuromuscular functions (Larish et al., 1988; Martin et al., 1992).

Several biomechanical changes arise during aging, although it is hard to describe a general walking pattern for 'the elderly', as individual differences can explain some of the conflicting results reported in the literature (Winter, 1991). Stride length and walking speed decrease in the elderly (Larish et al., 1988; Martin et al., 1992; Winter et al., 1990). In the mechanics, a reduced push-off is seen (DeVita and Hortobagyi, 2000; Winter et al., 1990) and plantar flexor weakness and reduced ankle power generation during the push-off are suggested to be important determinants of reduced walking speed in the elderly (Bohannon, 1997; Kerrigan et al., 1998; Winter et al., 1990). Some changes could be related to adopting a safer gait or are related to reduced dynamic balance (Winter et al., 1990).

Also in several (gait) pathologies, the inability to walk efficiently because of locomotion pathologies leads to an increase in metabolic energy expenditure. For some clinical populations, the metabolic energy expenditure is thereby up to twice the normal energy expenditure (Waters and Mulroy, 1999). Part of this can be due to the inability to use elastic storage and recoil, because of reduced coordination of plantar flexor muscles or because of plantar flexor weakness, e.g. in patients with stroke, spinal cord injury, cerebral palsy or in elderly (Sawicki et al., 2009). Also, redistribution of joint work to compensate for the reduced joint work due to a pathology, e.g. in the ankle, often comes with an increased metabolic cost (Sawicki et al., 2009; Wutzke et al., 2012).

With age or in specific pathologies, the walking pattern shows changes in the mechanics and energetics that reduce the efficiency and increase the metabolic energy cost.

3. Exoskeletons: shift in focus

Despite the fact that exoskeletons are not (yet) part of our daily lives, several attempts were done to build exoskeletons to assist human locomotion since the patent of Nicholas Yagn (Yagn, 1890). We will only discuss exoskeletons on which information is published on the design, control and/or performance of the device. Several devices are described in the literature (Fig. 10) to improve performance of healthy subjects (e.g. Jacobsen et al., 2004; Kazerooni & Steger, 2006; Kazerooni, 2005; Pratt et al., 2004; Zoss et al., 2006) or restore performance of impaired subjects during walking (e.g. Belforte et al., 2001; Hughes, 1972; Ruthenberg et al., 1997; Vukobratovic et al., 1974). While these seemed promising, several technical problems, weight and battery limitations, or control issues limited the practical use (Hughes, 1972; Jacobsen et al., 2004; Pratt et al., 2004). Also, walking efficiency was often not measured or showed that these exoskeleton increased the metabolic energy cost (Gregorczyk, 2006; Walsh et al., 2007). None of these complex devices was able to reduce the metabolic energy cost of walking under normal conditions, meaning that normal walking without exoskeleton was always easier compared to walking with the exoskeleton. Insufficient attention on human movement biomechanics and physiology might be the main cause of the rather disappointing results (Ferris et al., 2007).

Fig. 10
Examples of several full body exoskeletons that were experimentally tested. From left to right: the powered gait orthosis of Ruthenberg et al. (Ruthenberg et al., 1997), the pneumatic active gait orthosis of Belforte et al. (Belforte et al., 2001), the powered exoskeleton for load carrying from Zoss et al. (Zoss et al., 2006) and the quasi-passive exoskeleton for load carrying augmentation of Walsh et al (Walsh et al., 2007).

Therefore, it was important to take a step back and apart from building exoskeletons, start to focus on the human-exoskeleton interaction. Walking is the result of a complex chain and changes in the environment (e.g. with speed, grade or carried weight) or in the gait pattern (e.g. with age or pathologies) influence the walking efficiency. Interaction of an exoskeleton with the highly efficient human system is difficult and can only be successful when assistance comes at the right timing, with just enough power, for just the necessary joints.

The primary outcome to evaluate the human-exoskeleton interaction in this thesis is the metabolic energy cost as our main goal is to reduce the metabolic cost of walking by improving the human-exoskeleton interaction. Besides that this would largely increase the number of possible applications for exoskeletons, this could also be considered a fundamental challenge, as the walking pattern is already energetically and mechanically optimized through evolution. Exoskeleton effectiveness could then be evaluated based on the metabolic energy cost of walking with the exoskeleton and the main goal is to reduce the metabolic cost of exoskeleton walking below the level of normal walking without exoskeleton. Measuring kinematics, kinetics, EMG and perception can then help to understand why the exoskeleton assistance is or is not successful and how the human-exoskeleton interaction can be improved. These measures are considered the secondary outcomes in this thesis.

Apart from the complex control and actuation, that was often not optimal for human walking, the large full-body exoskeletons that were built in the beginning of this century often consisted of a heavy construction and a large power source, which adds a lot of weight and largely increases the metabolic energy cost of walking. Therefore, smaller exoskeletons that focus on one joint gained more interest in the last decade. Smaller exoskeletons that are focussed on one joint are also easier to control and the effects on primary and secondary outcomes are easier to attribute to the assistance of that specific joint. One aspect of human locomotion – the importance of ankle plantar flexion during the push-off – lead to increased attention towards ankle-foot exoskeletons that assist plantar flexion (Fig. 11). These were initially introduced as a tool to study fundamentals of locomotion (Ferris, Czerniecki, et al., 2005; Ferris, Gordon, et al., 2006) but soon showed to have indeed potential for reducing the metabolic cost of walking and restoring gait impairments. This thesis focusses on such ankle-foot exoskeleton that assist plantar flexion during walking.

We will not focus on exoskeletons that act in series with a user's limb (Dollar and Herr, 2008) and thereby influence the forces that are applied on the body as they often include prolonged legs and are limited to be used for persons with gait problems. Exoskeletons for paraplegic users like the ReWalk (Esquenazi et al., 2012) or Ekso suit (Strickland, 2012) will also not be discussed. For these patients, any improvement leading to an (even not optimal) walking pattern is a success while in healthy subjects or subjects with reduced walking abilities, the interaction between the human system and the exoskeleton itself needs to be optimal to be successful. It is that complex interaction, between the human dynamical system and the exoskeleton dynamical system, that is the focus of this thesis.

Although the ultimate goal for exoskeletons is to assist persons with mobility problems, the core of this thesis is on healthy subjects. It is important to first study exoskeleton assistance in healthy subjects to fully understand how exoskeletons interact with locomotion before pathological gait can be tackled.

Otherwise, the risk of reducing walking abilities instead of improving them and the risk of compensation mechanisms in other joints is too high. Focusing on healthy subjects as a first step allows to gather general information that can be used for applications in several pathologies, next to applications for healthy subjects.

This thesis focusses on the human-exoskeleton interaction for plantar flexion assisting exoskeletons in healthy subjects to reduce the metabolic cost of exoskeleton walking.

4. First ankle-foot exoskeletons

In order to situate the research that is done in this thesis within the field of research on ankle-foot exoskeletons, the next pages will describe exoskeleton research with ankle-foot exoskeletons in the last decade until the time we started our research. As developments in the field of exoskeletons are going fast, this allows to understand the framework that lead to our research questions. In the discussion section, more recent developments and future perspectives will be discussed in relation to our findings. The focus will be on sagittal plane movements because these ankle-foot exoskeletons are supposed to assist mainly in the plane of progression.

Initial research on adaptations during walking with lower limb exoskeletons did not focus on the ankle joint (Dietz et al., 2004; Hesse et al., 2003) and the first powered ankle-foot orthoses aimed to study the effects of well-defined perturbations (Andersen and Sinkjaer, 1995, 2003) or assisting drop-foot gait (Blaya and Herr, 2004) but were only used for dorsiflexion assistance. The first powered ankle-foot orthosis that provided plantar flexion power (Fig. 11) was built to study the relationship between biological mechanical work and metabolic energy cost during walking and was believed to have potential for locomotion studies or gait rehabilitation (Ferris, Czerniecki, et al., 2005).

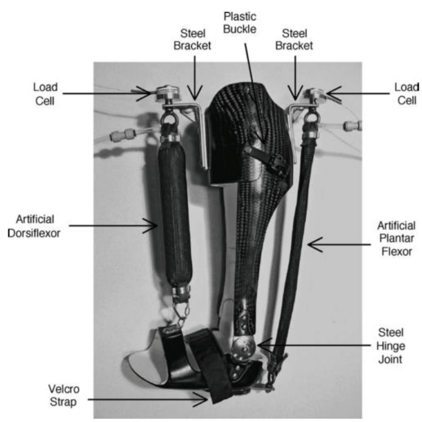

Fig. 11

An improved version of the first exoskeleton of Ferris et al. (Ferris, Czerniecki, et al., 2005), with pneumatic muscles for both plantar flexion and dorsiflexion assistance. This exoskeleton can be worn over the foot and the lower leg. From Ferris et al. (Ferris, Gordon, et al., 2006).

The lightweight exoskeleton (Fig. 11) of Ferris et al. (Ferris, Czerniecki, et al., 2005) (1.6 kg per exoskeleton), consisting of an ankle-foot orthosis with pneumatic muscles, was the basis for the exoskeletons that were used throughout this thesis (which are described in detail in *chapter 1*). These pneumatic muscles share some of the analogies of the human system by a simple design, consisting of a rubber tube covered with a braided shell, invented by Joseph McKibben in the 1950s (Tondu and Lopez, 2000) and regained interest with exoskeleton development. Pneumatic muscles mimic the contractile element of biological muscles (Klute et al., 2002). When compressed air is pressed into the rubber tube, it expands and the braided shell constrains the expansion and causes shortening of the pneumatic muscle. These pneumatic muscles are useful in exoskeletons because of the relatively low weight and the high forces that they can cause while still remaining compliant (Klute et al., 2002). They were not only used in ankle-foot exoskeletons but also in knee (Sawicki and Ferris, 2009b) and hip exoskeletons (Do Nascimento et al., 2008), although these attempts were so far less successful. While bench-top testing shows a linear force-length relationship, human experiments are necessary to study the behaviour of these pneumatic muscles during walking as the pneumatic muscle force is influenced by the muscle length, which results in a non-linear relationship between the control signal and the pneumatic muscle force during walking with these exoskeletons powered with pneumatic muscles (Gordon et al., 2006).

This thesis focusses on ankle-foot exoskeletons powered with pneumatic muscles that contract when they are inflated with compressed air and assist plantar flexion.

5. Exoskeletons as an experimental tool

Initial research with pneumatic ankle-foot exoskeletons was mainly focussed on studying neuromechanical adaptations during walking with unilateral exoskeletons. Exoskeleton assistance was evaluated based on joint kinematics and lower limb EMG by comparing powered exoskeleton walking with unpowered exoskeleton walking (which is walking with the exoskeleton without pneumatic muscle assistance). Research focussed on how subjects walked with the exoskeleton and if they were able to reduce biological muscle activity of the plantar flexor muscles as a result of the plantar flexor assistance of the exoskeleton. As a control method a proportional pneumatic muscle controller was often used (Fig. 12), in which exoskeleton assistance followed the muscle activity pattern of the biological soleus muscle (Cain et al., 2007; Ferris, Czerniecki, et al., 2005; Ferris, Gordon, et al., 2006; Gordon and Ferris, 2007; Kao et al., 2010) or the gastrocnemius muscle (Kinnaird and Ferris, 2009). Others used an on-off controller (Fig. 12), where assistance was simply switched on or switched off based on kinematic properties (Cain et al., 2007; Gordon et al., 2006). Due to the non-linear properties of the pneumatic muscles in combination with the human ankle behaviour, the torque pattern of the exoskeleton was close to that of the ankle plantar flexors in normal human walking.

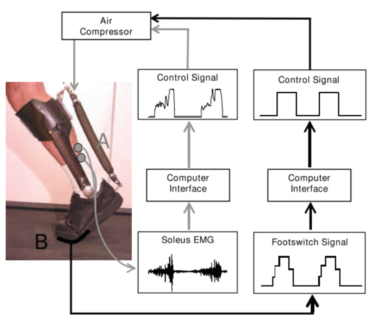

Fig. 12
Representation of the proportional EMG control method used in most unilateral exoskeleton studies (A) in grey on the left side and the kinematic control system (B), which is a simple on-off controller where actuation is on or off based on kinematic properties. From Cain et al. (Cain et al., 2007).

In these unilateral exoskeleton studies, plantar flexion assistance mainly influenced the ankle joint, while the knee and hip joint kinematics remained more or less normal during walking (Cain et al., 2007; Ferris, Czerniecki, et al., 2005; Gordon and Ferris, 2007; Gordon et al., 2006; Kinnaird and Ferris, 2009). In most studies, plantar flexion assistance resulted in reduced dorsiflexion during stance and earlier plantar flexion onset in the push-off. During walking with the proportional EMG controlled exoskeleton, where the pneumatic muscle follows the biological soleus or gastrocnemius muscle, the most important effect of plantar flexion assistance was the reduction in soleus muscle activity with up to 50 % compared to unpowered exoskeleton walking (Cain et al., 2007; Ferris, Czerniecki, et al., 2005; Gordon and Ferris, 2007; Kinnaird and Ferris, 2009). Some studies also reported reductions in medial and lateral gastrocnemius muscle activity (Cain et al., 2007; Kinnaird and Ferris, 2009). These reductions in muscular activity were promising for applications in impaired subjects and suggested a reduction in metabolic energy cost as muscle activity requires metabolic energy use.

In most studies, it took some time to learn to walk with the exoskeleton. Ankle joint angles and muscle activity patterns were strongly perturbed in the first minutes of plantar flexion assistance but after a 30 min adaptation these became closer to normal (Cain et al., 2007; Gordon and Ferris, 2007). Studying the adaptation showed that subjects needed between 20 and 25 min before ankle kinematics, soleus and gastrocnemius muscle EMG or orthosis work were adapted to the plantar flexion assistance. Adaptation was evaluated based on when kinematics, EMG or orthosis parameters where close to the values of these parameters at the end of two 30 min sessions. These studies showed that subjects could use the newly learned task almost immediately on a second day (Cain et al., 2007; Gordon and Ferris, 2007; Kinnaird and Ferris, 2009). Stronger exoskeleton assistance, by adding two pneumatic muscles to the exoskeleton instead of one resulted in longer adaptation times (Kao et al., 2010). This suggests that adaptation time is dependent on the amount of ankle joint mechanical power that is delivered by the exoskeleton. Although the proportional EMG control method directly influences the relationship between human muscle coordination and ankle joint dynamics, adaptation rates were similar compared to kinematic on-off control based on forefoot contact (Fig. 12)(Cain et al., 2007). The EMG control method was favoured (1) due to kinematics that were closer to normal walking, (2) because it seems a more natural control mechanism and (3) because of larger reductions in soleus EMG, potentially leading to larger reductions in metabolic cost. However, metabolic cost of walking was not measured and recent research showed that kinematic control based on forefoot contact is not optimal (Malcolm et al., 2013). Less attention was given to the higher positive orthosis work (which may induce larger reduction in metabolic energy cost) and the fact that ankle angle and pneumatic muscle torque patterns seemed to change less over time in the footswitch control method (which suggests that it is easier to learn to walk with the exoskeleton). Therefore, the choice of control method should depend on the goal of the exoskeleton.

Initial research with unilateral plantar flexion assisting exoskeletons showed that exoskeleton assistance could lead to a reduction in muscle activity of the plantar flexors.

6. The road to beating the human system

Unilateral exoskeleton experiments increased understanding of the human-exoskeleton interaction and confirmed the potential of exoskeletons, especially because of the reduction in soleus muscle activity. Still, bilateral assistance seemed a necessity in order to reduce the metabolic energy expenditure of walking with ankle-foot exoskeletons. With the first bilateral exoskeleton studies, metabolic energy cost was effectively measured to study the effect of exoskeleton assistance on metabolic energy consumption.

The first experiments with bilateral exoskeletons used a kinematic on-off controller that resulted in assisting plantar flexion during the push-off (Norris, Granata, et al., 2007; Norris, Marsh, et al., 2007). These studies indicated that after only a few minutes of adaptation, subjects could reduce the metabolic cost of transport with 15% compared to unpowered exoskeleton walking. Also, plantar flexion assistance with an exoskeleton increased the preferred walking speed, even when compared with standard shoes (Norris, Granata, et al., 2007). This demonstrated that the human neuromuscular system can adapt to an external energy source and can use it to decrease the metabolic energy expenditure during bilateral exoskeleton walking. The assistance of the exoskeleton was sufficient to compensate for the hindrance and the weight of the device but did not reduce the metabolic cost when compared to normal walking (with normal shoes and thus without an exoskeleton).

Sawicki et al. also used bilateral ankle-foot exoskeletons with an EMG-controlled exoskeleton to study the role of the ankle plantar flexor muscles during the push-off in walking (Sawicki and Ferris, 2008, 2009c). After app. 90 min of adaptation, assistance of the exoskeleton only influenced ankle kinematics, with earlier and increased plantar flexion during the push-off and caused reductions in soleus EMG. The bilateral assistance required longer adaptation compared to unilateral assistance. Probably as a result of the reduced soleus EMG, exoskeleton assistance lead to a reduction in net metabolic cost of 10% compared to unpowered walking (Sawicki and Ferris, 2008, 2009c). Again, no differences were found when compared to normal walking at normal walking speeds (Sawicki and Ferris, 2009c). With the same exoskeleton they also studied uphill walking with an exoskeleton and reported reductions in the metabolic cost of 10 to 13% compared to unpowered exoskeleton walking, depending on the inclination (Sawicki and Ferris, 2009a).

The main goal in these studies was not to find the highest reductions in metabolic cost but mainly to use the exoskeleton as a tool for fundamental neuromechanical research to study how subjects adapt to an exoskeleton that replaces part of the biological ankle work during the push-off. Therefore, Sawicki et al. introduced the ‘Exoskeleton performance index’ and the ‘Ankle joint apparent efficiency’ (Sawicki and Ferris, 2008).

$$Exoskeleton\ performance\ index = \frac{\Delta\ Net\ metabolic\ power\ \times\ \eta^{+}_{muscle}}{\begin{array}{c}Average\ exoskeleton\\ positive\ mechanical\ power\end{array}}$$

The estimated efficiency of human muscles that perform positive work, η^{+}_{muscle}, is estimated to be around 0.25 based on experimental work during steep uphill inclines (Margaria, 1968). This means that performing 1 J of mechanical work consumes about 4 J of metabolic energy. Assuming that the exoskeleton replaces biological ankle work, the exoskeleton performance index could be seen as an upper bound of the fraction of ankle joint positive mechanical work performed by active muscle shortening (Sawicki and Ferris, 2008). A performance index of 1 indicates that 100% of the exoskeleton work replaced biological muscle work, a lower performance index indicates that the exoskeleton work replaced biological muscle work and elastic energy recoil.

$$Ankle\ joint\ apparent\ efficiency = \frac{\begin{array}{c}Average\ exoskeleton\\ positive\ mechanical\ power\end{array}}{\Delta\ Net\ metabolic\ power} = \frac{\eta^{+}_{muscle}}{\begin{array}{c}Exoskeleton\\ performance\ index\end{array}}$$

The ankle joint apparent efficiency could be seen as an indication of the amount of elastic recoil in the ankle. An apparent efficiency of 0.25 would indicate that all of the positive work in the ankle is performed by active muscle shortening, a higher apparent efficiency again points towards elastic recoil. Sawicki et al. showed that elastic recoil accounts for around 60% of the positive work during the push-off in walking (Sawicki and Ferris, 2008), which reduces during walking at faster speeds with longer steps (Sawicki and Ferris, 2009c) and also reduces with inclination during uphill walking (Sawicki and Ferris, 2009a). They showed that while the plantar flexor muscle-tendon performed around 35% of the total positive mechanical work of the lower limbs during level walking, they only consumed around 20% of the metabolic energy (Sawicki and Ferris, 2008, 2009c). This corresponds with results of modelling studies that suggest that the ankle extensors consume around 25% of the metabolic energy of walking (Umberger and Rubenson, 2011). This suggests that plantar flexion assistance, by replacing positive joint work in the ankle with mechanical work of the exoskeleton, could maximally reduce the metabolic cost with around 20%. It is therefore important to not only focus on joints that generate most mechanical power but also keep in mind the amount of recycled tendon elastic energy that contributes to the mechanical power (Sawicki and Ferris, 2009a).

Initial research with bilateral exoskeletons emphasized the possibility to reduce metabolic cost of walking. However, no differences with normal walking were (yet) found.

7. Optimizing exoskeleton assistance enables human walking

Until 2013, no exoskeleton study was able to show a reduction in the metabolic cost of walking for exoskeleton walking compared to normal walking. Based on previous studies (Norris, Granata, et al., 2007; Sawicki and Ferris, 2008, 2009c) this should be possible if (1) the penalty of wearing the exoskeleton could be reduced or (2) if exoskeleton assistance could be optimized in a way that larger reductions in metabolic cost are realized. Although most exoskeleton studies focussed on an EMG-controlled exoskeleton, this does not allow to modify exoskeleton assistance easily. While it seems natural to apply plantar flexion power in accordance with the biological muscle activity of the plantar flexors, it was unclear if this is optimal to reduce the metabolic cost of walking. Also, while unilateral exoskeleton studies found some benefits in the proportional EMG control (Cain et al., 2007), the bilateral exoskeleton studies that resulted in the largest reductions in metabolic energy cost used a kinematic on-off controller (Norris, Granata, et al., 2007). Additionally, adaptation times in these studies (less than 20 min) were shorter compared to adaptation times of EMG controlled exoskeleton studies (up to 90 min) (Sawicki and Ferris, 2008, 2009c).

Several exoskeleton assistance parameters could have an influence on the metabolic cost of walking (e.g. the contraction speed, the assistive torque, the joint angle, etc.). The exoskeleton power curve (Fig. 13B), which is the product of the exoskeleton torque and the ankle joint angular velocity, is a measure that combines exoskeleton dynamics and human dynamics and therefore seems a good parameter to optimize exoskeleton assistance. More details on this exoskeleton power are described in *chapter* 1. Two characteristics of the exoskeleton power curve, the actuation onset timing (= when does the exoskeleton start to assist) and the average amount of exoskeleton power (= how much assistance does the exoskeleton deliver), can be controlled with an exoskeleton with kinematic control and intuitively seems parameters that influence the metabolic cost of walking.

As a first study focussing on optimizing exoskeleton actuation, our lab used a footswitch controlled exoskeleton where actuation onset and ending of the pneumatic muscles could be set based on heel contact timing in order to study the effect of actuation timing on the metabolic energy cost of exoskeleton walking (Malcolm et al., 2013). This exoskeleton was called WALL-X (Wearable Assistive Lower Leg eXoskeleton). When pneumatic muscles were actuated between 43 and 63% of the stride, exoskeleton assistance resulted in a maximal reduction in metabolic cost of 17% versus unpowered walking (Fig. 13A) and a performance index larger than 1. For the first time this also resulted in a reduction versus normal walking of 6%, despite the additional weight and the disturbance of the walking pattern. A U-shaped pattern was found for the relationship between start of actuation of the pneumatic muscles and the metabolic cost of walking, with an optimum around 40% of the stride time. Earlier and later actuation onset timings resulted in a smaller reduction in the metabolic cost of walking. Our results showed that plantar flexion assistance resulted in a more effectively redirection of the bCOM during the step-to-step transition, which suggests that plantar flexion assistance not only replaces positive ankle joint work but also has an influence on overall walking mechanics and energetics.

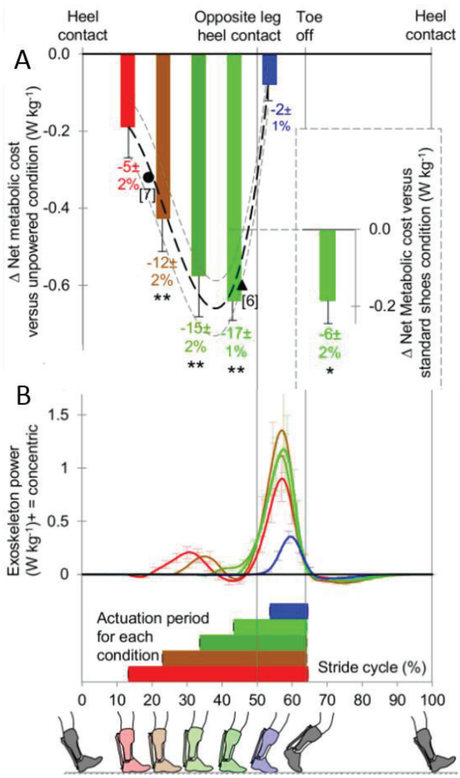

Fig. 13

Reductions in metabolic cost (expressed as the reduction versus the unpowered condition) during walking with WALL-X (A) for several powered exoskeleton conditions with varying actuation onset timings (B). The coloured vertical bars represent the reduction in metabolic cost for the powered conditions. The actuation durations of the pneumatic muscles are represented with horizontal coloured bars at the bottom of the figure and result in different exoskeleton power profiles for each condition. [5] and [6] are result of the studies of Norris et al. (Norris, Granata, et al., 2007) and Sawicki et al. (Sawicki and Ferris, 2008). Adapted from Malcolm et al. (Malcolm et al., 2013).

Although the optimal actuation timing started much later compared to the biological soleus muscle, it is not surprising that this condition was optimal, as it coincided with the onset of positive power in the ankle joint during walking (Winter, 1983). In the beginning of the stance phase the ankle plantar flexors delivers negative work due to the eccentric action in controlling passive ankle dorsiflexion and it seems that additional power of the exoskeleton is less efficient during this power absorption phase. When reductions in metabolic cost and actuation timings of other studies (Norris, Granata, et al., 2007; Sawicki and Ferris, 2008) were added, they seemed to match with our results (Fig. 13A). This indicates that actuation timing is an important determinant for the reduction in metabolic cost. Smaller reductions in metabolic cost in other studies could (at least partly) be attributed to a not optimal actuation timing.

The amount of average exoskeleton mechanical power over a stride is also believed to influence the metabolic cost of walking and can be controlled by adjusting the air pressure of the pneumatic muscles. In different exoskeleton studies that reported reductions in net metabolic cost during level walking, average positive exoskeleton ankle power (summed for both legs) ranged from 0.02 to 0.24 $W \cdot kg^{-1}$ and resulted in reductions in metabolic cost of 2 to 17% compared to unpowered walking (Malcolm et al., 2013; Norris, Granata, et al., 2007; Sawicki and Ferris, 2008, 2009c). During unpowered exoskeleton walking, total average positive ankle joint power over a stride is around 0.38 W/kg, with around 0.19 W/kg for each ankle (Sawicki and Ferris, 2008, 2009c). In a unilateral exoskeleton study, Jackson et al. (Jackson and Collins, 2014) showed that the metabolic cost decreases exponentially when exoskeleton mechanical power was increased until about twice the biological ankle joint mechanical work during walking. Although it seems evident that with low amounts of exoskeleton power the effect on metabolic cost will be small and that with too much exoskeleton power it will be hard to keep on walking comfortably, the precise relationship between the amount of exoskeleton mechanical power and the metabolic cost of bilateral exoskeleton walking is unclear as it is hard to translate the results of the unilateral study to walking with WALL-X.

Our timing study shows that it is hard to estimate optimal exoskeleton parameters based on reference data of normal walking, especially as the assistive mechanism is complex and even seems to influence bCOM dynamics. The results of Farris et al. in hopping with an elastic exoskeleton, where they showed that this influenced muscle fascicle lengths (Farris et al., 2013), emphasizes the complexity of interacting with human locomotion, and the possible risk for negative compensatory effects. Therefore, the approach where metabolic reductions are optimized by exploring human responses to changes in different parameters of the actuation, like we did in our timing study, is extremely useful. By measuring secondary outcomes this also helps to understand how human walking is influenced by exoskeleton assistance.

Our lab was the first to report a reduction in metabolic cost of walking with an exoskeleton compared to normal walking by optimizing exoskeleton actuation timing.

8. Practical use of exoskeletons

Since reductions versus normal walking were shown (Malcolm et al., 2013), ankle-foot exoskeleton can become a tool to assist impaired people during rehabilitation or in daily life, by improving the walking pattern or by reducing the metabolic cost of walking, which is often increased in gait pathologies. While applications for impaired subjects seem evident, healthy subjects could benefit from an exoskeleton to reduce the metabolic cost during hill walking, while carrying weight or to increase performance (in terms of speed, duration or carried weight), while still maintaining the flexibility of walking. This could lead to military applications, applications for firefighters or construction workers, or even tools for assisting load carrying or uphill walking during leisure activities.

One of the important goals in gait rehabilitation, which could be achieved with exoskeleton assistance, is the recovery of mechanisms that reduce metabolic cost and increase stability (Kuo and Donelan, 2010). To do so, an exoskeleton needs to be optimized for the specific needs of the population. As long as the assistive mechanism is not entirely clear, caution is necessary in the applications of exoskeletons in subjects with specific gait pathologies, especially when they have muscle or bone weakness. A reduction in the metabolic cost of walking with exoskeleton assistance seems positive in subjects with an increased metabolic cost of walking but it does not exclude negative influences on other muscles or joints. It is possible that higher demands on some physiological structures are necessary during exoskeleton walking, which might be not favourable. During hopping with an elastic exoskeleton, research showed that muscle activity and force in the soleus are reduced and that metabolic cost reduces with almost 20% (Farris et al., 2013). However, fascicle power output was unchanged and muscle fascicle lengths were even increased. This might be problematic for subjects with specific gait pathologies, especially because excessive muscle strain is related to muscular injuries (Lieber and Fridén, 1993). This highlights the complexity of the human-exoskeleton interaction and the importance of studying the human-exoskeleton interaction, on a mechanical, energetic and muscular level in order to eliminate negative compensatory effects of exoskeleton assistance.

Exoskeletons are often seen as a tool for rehabilitation after neurological injury as subjects need specific practice with a high degree of participation and exoskeletons could replace manual assistance of the therapists (Collins and Jackson, 2013; Ferris, Sawicki, et al., 2005). Stroke survivors and patients with spinal cord injury are increasing in number in the general population (Truelsen et al., 2006) and often experience neurological deficits, the inability to walk unassisted and reduced ankle push-off (Jonkers et al., 2009). The latter plays an important role in gait asymmetry, compensatory mechanisms, reduced walking speed and increased metabolic cost of walking (Nadeau et al., 1999; Stoquart et al., 2012), which could be improved with exoskeleton assistance. Several walking machines are used to perform locomotion practice (e.g. Colombo et al., 2000). However, plantar flexion was often not included in previous devices (Colombo et al., 2000; Ferris, Sawicki, et al., 2005; Hesse et al., 2003; Reinkensmeyer et al., 2004) but recent devices include ankle joint guidance which is useful as the ankle joint is critical for mobility after a neurological injury (Kim et al., 2013; Nadeau et al., 1999). Clear evidence that robot-assisted gait training improves walking function more than other training strategies is not present (Duerinck et al., 2012; Hidler et al., 2009; Swinnen et al., 2010) although ankle-foot exoskeletons have shown to improve ankle push-off kinematics in patients with incomplete spinal cord injury (Sawicki et al., 2006), to reduce the number of drop foot events and increase symmetry during impaired locomotion (Blaya and Herr, 2004) and could be used to drive gait patterns (Collins and Jackson, 2013). A drawback for ankle-foot exoskeletons as a rehabilitation or gait restoring tool might be the reduced stability that goes with plantar flexion assistance (Norris, Marsh, et al., 2007) especially as active stabilization during walking comes with a cost (Donelan et al., 2004). This can be problematic as the target population will probably have pre-existing musculoskeletal or other neurological limitations that may impair the ability to compensate for mechanically induced instabilities.

Another possible target population are the elderly. With age the maximal aerobic power and physical capabilities reduce, which may lead to limited performance in daily life activities. This may result in small aerobic reserves during walking, which is related to the reduced walking speed in elderly (Schrack et al., 2010). Exoskeletons could, especially in the elderly that suffer from reduced exercise tolerance, increase performance while still maintaining a comfortable aerobic reserve. Norris et al. (Norris, Granata, et al., 2007) showed that there is potential for exoskeleton assistance in elderly. Mobility, physical performance, of which walking ability and gait speed are important measures, and physical activity are all predictors of disability and mortality (Guralnik et al., 2000; Lee, S. J. et al., 2006; Rosano et al., 2008; Studenski et al., 2011; Tsai et al., 2007). This become especially important as the number of older people is growing and will continue growing in the next decades. Performing physical activity, also in the form of walking (Manson et al., 2002), can improve cardiorespiratory fitness, also in sedentary elderly (Kohrt et al., 1991) and reduces mortality risk (Blair et al., 1995). However, due to mobility problems, reduced walking speed, reduced physical fitness or physical constrains, some elderly are unable to meet the aerobic physical activity recommendations for elderly (Nelson et al., 2007). Exoskeleton could help them to reach the threshold of 1000kcal/week, leading to 20-30% reduction in risk of all-cause mortality (Lee, I. M. and Skerrett, 2001) by improving walking performance and mobility. This concept would be similar to the use of a power assisted bicycle (Louis et al., 2012). However, several physiological and biomechanical changes arise with age, which should be kept in mind to successfully assist gait in elderly.

Another unexplored possibility is to use an ankle-foot exoskeleton in other patients with reduced exercise tolerance in general, or patients with Chronic Obstructive Pulmonary Disease (COPD) in specific. COPD, which is the 4th most important cause of mortality worldwide (Shahab et al., 2006), is characterized by airflow limitation due to smoking, leading to shortness of breath and reduced exercise tolerance. Since the lung damage is irreversible (Mcalister et al., 2003), the room for improvement by traditional rehabilitation therapy is limited. Walking assistance by means of an exoskeleton in COPD patients, who typically have to rest after a few minutes of walking, can allow longer uninterrupted walking bouts at a lower but maintainable metabolic cost and with less dyspnea. If effectiveness of ankle-foot exoskeleton can be showed in this population, this opens perspectives towards the development of ankle-foot exoskeletons as an assistive device in daily life. Also, if walking with an exoskeleton is included in the rehabilitation, the longer walking bouts or higher walking speeds during the rehabilitation sessions can have a positive impact on rehabilitation outcomes such as lower limb strength, submaximal exercise capacity (6 min walking distance) and quality of life. While the application of exoskeletons in these patients seems evident, the short and discontinuous walking effort implies control and design challenges for the exoskeleton.

Several populations could benefit from exoskeleton assistance. A rather unexplored option is the use of exoskeletons in patients with reduced exercise tolerance.

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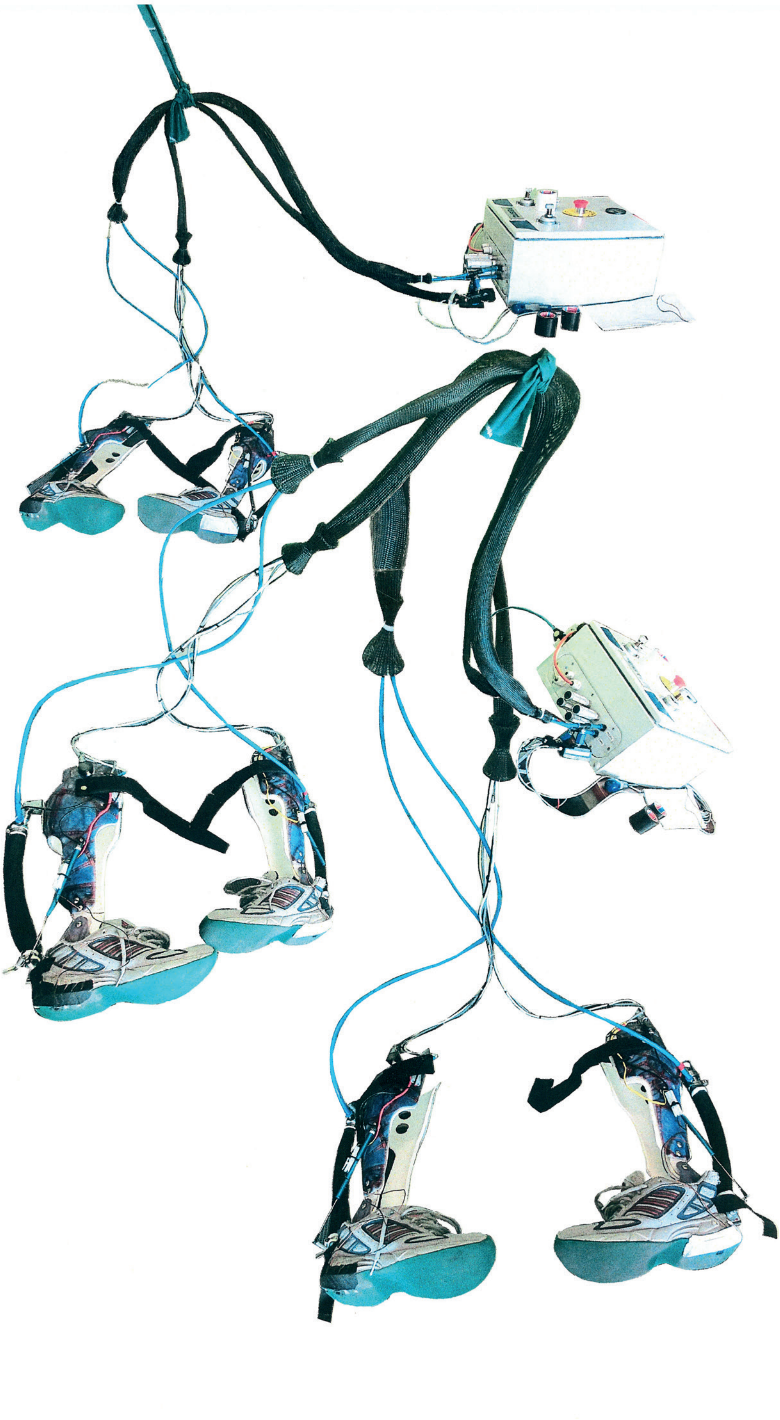
RESEARCH QUESTIONS



RESEARCH QUESTIONS

1. Overview

"You think you know what a person needs, but you don't.
The person is complex, tricky and adapts in strange ways" (Collins, 2014)

The mission of the exoskeleton lab at the Ghent University is to enable the rise of a new generation of exoskeletons that augment mobility to a clearly perceivable degree, in able-bodied as well as impaired persons, by optimizing assistance patterns based on evidence from human neuromechanics experiments.

In the introduction it was shown that walking is a complex movement and that previous exoskeletons where unsuccessful in reducing the metabolic cost of walking due to a not optimal interaction between human walking and exoskeleton assistance. Our goal is to focus on the human-exoskeleton interaction in order to reduce the metabolic cost of walking. By improving the understanding of the human-exoskeleton interaction we want to use exoskeletons for practical applications in healthy and impaired subjects.

Smaller ankle-foot exoskeletons allowed to focus on assisting the ankle, where the largest joint work is delivered during walking. This also allowed to study the human-exoskeleton interaction more directly and to optimize exoskeleton assistance. Due to the complexity of walking, it is hard to estimate the human-exoskeleton interaction based on reference data of normal walking. Therefore, a systematic approach where metabolic reductions are optimized by exploring human responses to changes in different parameters of exoskeleton actuation is suggested to find the largest reductions in metabolic cost. This also allows to study why and how human responses change with changes in exoskeleton actuation parameters. By doing so, Malcolm et al. (Malcolm et al., 2013) were able to reduce the metabolic energy cost of exoskeleton walking below that of normal walking. We aim that our findings can advance exoskeleton research and can be used by others to improve exoskeletons so that one day subjects can benefit from exoskeleton assistance during ambulant walking. The metabolic cost of walking will always be our primary outcome. Secondary outcomes like walking kinematics, EMG and exoskeleton kinetics will be used to explain the reduction in metabolic cost and to improve our understanding of the human-exoskeleton interaction and the assistive mechanism.

A common problem in exoskeleton research is the variety of exoskeleton designs and control algorithms, which makes it hard to compare findings between studies. Therefore, we described our exoskeleton, which we called WALL-X (Wearable Assistive Lower Leg eXoskeleton), and the control mechanism in *chapter* 1 with attention to the improvements during the time-course of this thesis. The technical improvements in *chapter* 1 allowed to focus on adaptation to walking with an exoskeleton (*chapter 2*) and optimizing exoskeleton actuation (*chapter* 3 and *chapter* 4) in order to use WALL-X to increase maximal performance (*chapter* 5) and as an application to assist elderly during walking (*chapter 6*). While each of the topics aims to answer a fundamental or practical research question, all studies contribute to overall understanding of the human-exoskeleton interaction.

2. Adaptation

Adaptation is defined as the time to reach a minimum in the metabolic cost of walking with an exoskeleton.

The most successful ankle-foot exoskeleton studies, in terms of the metabolic energy cost, reported reductions in metabolic energy cost of 15 to 17% when compared with unpowered walking (Malcolm et al., 2013; Norris, Granata, et al., 2007), or 6% when compared with normal walking (Malcolm et al., 2013). These studies used a relatively simple on-off controller based on kinematic parameters and the metabolic reductions were found after relatively short habituation to learn to walk with the exoskeleton (less than 15 min)(Malcolm et al., 2013; Norris, Granata, et al., 2007). As habituation times are substantially longer in studies with EMG controlled exoskeletons (up to 90 min)(Sawicki and Ferris, 2008, 2009c), we suggested that longer habituation to the on-off controlled exoskeleton could further reduce the metabolic cost of walking. Also, the adaptation process to walking with these exoskeletons was not yet described and it was unclear how long subjects needed to walk with these exoskeletons before they are metabolically adapted: i.e. when metabolic cost reaches a minimum. Studying the metabolic adaptation will allow to generate adaptation duration guidelines for simple exoskeletons that can be used in future experiments or in practical applications.

Most studies that previously focussed on adaptation to walking with a plantar flexion assisting exoskeleton used a unilateral exoskeleton (Cain et al., 2007; Gordon and Ferris, 2007; Gordon et al., 2006; Kao et al., 2010; Kinnaird and Ferris, 2009). The time to reach a steady state in walking kinematics, lower limb EMG patterns and exoskeleton dynamics was situated between 20 and 25 min. Adaptation times did not drastically differ between exoskeletons with porportional soleus EMG control (Gordon and Ferris, 2007), proportional gastrocnemius EMG control (Kinnaird and Ferris, 2009) or a kinematic-based controller (Cain et al., 2007). However, the results of unilateral walking cannot be used for bilateral walking, especially as the time to learn to walk with a bilateral EMG-controlled exoskeleton (up to 90 min) (Sawicki and Ferris, 2008, 2009c) is much longer compared to reported adaptation times for unilateral walking with a similar exoskeleton (20 to 25 min) (Cain et al., 2007; Gordon and Ferris, 2007; Kao et al., 2010; Kinnaird and Ferris, 2009). It is therefore still unclear how long adaptation to walking with a bilateral on-off controlled exoskeleton takes.

The most important limitation of the previous research however, is that none of the studies that focussed on adaptation measured metabolic energy cost. Adaptation times were mainly evaluated based on EMG, kinematics and exoskeleton kinetics. As metabolic cost is our prime measure and allows to evaluate exoskeleton effectiveness, we evaluated adaptation time based on the metabolic cost.

In order to further understand the substantial reductions in the metabolic cost of walking after short adaptation times during walking with WALL-X in a previous study (Malcolm et al., 2013), we used WALL-X to study how long metabolic adaptation takes during walking with bilateral exoskeletons in *chapter* 2. Kinematics, EMG and exoskeleton kinetics were used to understand how neuromechanical changes occurred during the metabolic adaptation. The reported adaptation times for kinematics, EMG and exoskeleton kinetics during unilateral walking with an EMG controlled exoskeleton of around 24 min (Gordon and Ferris, 2007) were used as the duration of our exoskeleton walking conditions. It was expected that subjects would metabolically adapt to exoskeleton walking within 24 min and that this would be accompanied by changes in kinematics, EMG and exoskeleton kinetics.

? How long does metabolic adaptation to walking with an exoskeleton takes?

? Which neuromechanical changes occur in the leg muscles during this adaptation?

3. Actuation

Actuation is defined as the assistance of the exoskeleton, characterized by the exoskeleton power curve. Although several parameters can be used to describe the assistance of the exoskeleton, we will focus on two key features of exoskeleton assistance: the actuation onset timing (when does the exoskeleton start to assist in the stride) and the average exoskeleton positive ankle mechanical power (how much assistance is delivered by the exoskeleton in a stride). Average exoskeleton positive ankle mechanical power is preferred over exoskeleton work as it takes into account differences in stride time and allows comparison with metabolic power.

The research of Malcolm et al. (Malcolm et al., 2013) showed that optimization of the actuation onset timing is crucial in order to find reductions in the metabolic cost of exoskeleton walking. We found a U-shaped pattern for actuation onset timing with an optimum around 40% of the stride. When other studies were compared, at least part of the differences in metabolic cost could be attributed to differences in actuation timing. It is likely that, besides other assistance parameters, also the average amount of positive exoskeleton ankle power over a stride is related to the metabolic cost of walking. When actuation timing was studied (Malcolm et al., 2013), average positive exoskeleton ankle power over a stride was not held constant, which could have influenced the relationship between actuation onset timing and metabolic cost of walking.

In *chapter* 3 we did a combined parameter sweep of actuation onset timing and average positive exoskeleton ankle mechanical power, in which one parameter was kept constant while the other was varied and vice versa. This allowed to study the relationship between actuation onset timing and metabolic cost and between average exoskeleton ankle mechanical power and metabolic cost, independent from each other. By performing a parameter sweep experiment we aimed to further optimize exoskeleton assistance, which could result in a larger reduction in the metabolic cost of exoskeleton walking. By applying a range of actuation timings and a range of average exoskeleton powers, while also measuring secondary outcomes (joint kinematics, EMG and exoskeleton kinetics), human-exoskeleton interaction could be better understood.

For timing a U-shaped pattern was expected based on our previous research (Malcolm et al., 2013). The relationship between average exoskeleton power and metabolic cost was currently untested for bilateral exoskeleton walking and based on the different exoskeleton studies that reported reductions in metabolic cost during level walking, it is hard to estimate the relationship between average positive exoskeleton ankle power and metabolic cost (Malcolm et al., 2013; Norris, Granata, et al., 2007; Sawicki and Ferris, 2008, 2009c). The highest reductions were found with the highest average exoskeleton power and an exponential relationship between average positive exoskeleton ankle power and metabolic cost was expected based on results in unilateral exoskeleton walking (Jackson and Collins, 2014).

?	**Can actuation onset timing be optimized (independent of average positive exoskeleton ankle mechanical power) in order to reduce the metabolic cost of exoskeleton walking?**

?	**Can average positive exoskeleton ankle mechanical power be optimized (independent of actuation onset timing) in order to reduce the metabolic cost of exoskeleton walking?**

To induce higher intensities in order to study maximal performance during exoskeleton walking in *chapter* 5, exoskeleton walking on steep uphill inclines seemed a good solution. However, exoskeleton assistance needs to be optimized for uphill walking to induce the highest reductions in metabolic cost. Due to changes in walking mechanics and energetics during walking on an inclination, we first studied the human-exoskeleton interaction during uphill walking. This also has relevance for applications for rescue workers, soldiers and in recreational hiking because walking is the most economic gait mode for steep uphill inclines (Ardigò et al., 2003).

Sawicki et al. (Sawicki and Ferris, 2009a) showed that an EMG-controlled exoskeleton could reduce the metabolic cost of uphill walking (on a 15% inclination) with 10% compared to unpowered exoskeleton walking. Due to the importance of the ankle joint during uphill walking, it is not surprising that a plantar flexion assisting exoskeleton can also reduce the metabolic cost of uphill walking. However, during level walking, the higher reductions in metabolic cost with an exoskeleton with optimized actuation timing (Malcolm et al., 2013) compared to an EMG controlled exoskeleton suggests that optimizing exoskeleton actuation during uphill walking could also result in higher reductions in metabolic cost. As mentioned before, the best way to optimize exoskeleton actuation timing is to explore the metabolic response to changes in actuation timing.

In *chapter 4* we studied the effect of actuation timing on the metabolic cost of uphill walking (on a 15% inclination). Kinematics, EMG and exoskeleton kinetics were measured to explain the reductions in metabolic cost during uphill walking and improve our understanding of the assistive mechanism. It was expected that an optimized actuation could reduce the metabolic cost of exoskeleton uphill walking with more than 10% compared to unpowered walking. Based on the effect of exoskeleton assistance on both plantar flexor muscle activity and bCOM dynamics during level walking with WALL-X (Malcolm et al., 2013), it was expected that exoskeleton assistance during uphill walking would also influence plantar flexor muscle activity and bCOM movements.

? What is the optimal actuation onset timing during uphill walking with an exoskeleton?

4. Performance

Performance is defined as the workload that can be performed for a certain effort. If an exoskeleton allows to walk faster or with more weight while the user has to perform the same effort, the exoskeleton increases performance. Maximal performance is defined as the workload that can be performed during a maximal effort. If an exoskeleton allows to walk faster or with more weight while the user performs a maximal effort, the exoskeleton increases the maximal performance.

We know that WALL-X can reduce the metabolic cost during steady-state level walking (Malcolm et al., 2013). This implies that exoskeletons can improve performance. For a certain metabolic cost, an exoskeleton allows to walk faster, on a steeper inclination or with more load. However, for practical applications in healthy subjects the goal is often to improve maximal (walking) performance. To achieve maximal performance during walking, a maximal effort is necessary and this implies higher walking intensities. These could be achieved by walking on an inclination or by carrying heavy loads.

We used WALL-X to study the effect of exoskeleton assistance on maximal walking performance in c*hapter* 5. To study the effect on maximal walking performance we used an incremental walking exercise test (Klimek and Klimek, 2007) to achieve maximal effort during walking. This method can be used to asses aerobic power during walking with an inclination of 15% by adding 5% of body weight every 3 min to a weight vest carried by the user. The goal was to study if WALL-X could improve the maximal walking performance during this maximal walking exercise test. However, this was only possible if the exoskeleton still reduced the metabolic cost during uphill walking with additional weight, also during higher intensities ánd if the user was still able to walk with the exoskeleton during uphill walking with additional weight, also during higher intensities. The maximal walking performance was expressed as the weight that subjects were carrying at volitional termination of the exercise test. It was hypothesized that exoskeleton assistance would still reduce the metabolic cost during uphill walking with additional weights and that subjects could reach maximal effort during uphill walking with additional weights with the exoskeleton. As a result, it was hypothesized that subjects could increase the maximal walking performance during exoskeleton assistance. It was expected that subjects increased the maximal carried weight at termination of the exercise test due to the lower effort for every weight because of the assistance of the exoskeleton. Physiological measures were used to verify if subjects could reach a maximal effort and to clarify how exoskeletons improve the maximal walking performance.

? Can an exoskeleton increase maximal walking performance?

5. Applications

Applications of exoskeletons are defined as situations in which the exoskeleton can be an added value to improve a certain practically relevant outcome.

Once exoskeletons are optimized and cause reductions in metabolic cost of more than 10% versus normal walking, the number of possible applications for exoskeletons increase. However, the relevance of reducing the metabolic cost of walking is sometimes questioned. Some people argue that this is a bad idea because reducing the metabolic energy expenditure could have a negative influence on health benefits of physical activity. However, walking assistance should be seen as an assistive tool for subjects with walking difficulties to incite them back on their feet. Especially those that suffer from reduced exercise tolerance or an increased metabolic cost because of age or because of specific pathologies like Chronic Obstructive Pulmonary Disease (COPD) could benefit from an assisting exoskeleton that reduces the metabolic energy cost of walking. This would allow them to improve participation in society and improve mobility, which will have a positive impact on physical activity and quality of life and can lead to associated health benefits. While this seems promising, more research is necessary to confirm this hypothesis. Also, research with an exoskeleton in hopping (Farris et al., 2013) showed the risk of negative effects on a muscular level because of exoskeleton assistance. More research with exoskeletons in specific populations is necessary before they can be used in a rehabilitation setting or in daily life to exclude possible negative compensatory effects of exoskeleton assistance.

A first step in conducting research with exoskeletons in specific populations is to study if these populations can indeed benefit from exoskeleton assistance and if exoskeleton assistance reduces the metabolic cost of walking. To explore the possible use of exoskeletons to assist elderly, *chapter 6* focusses on the effect of walking with WALL-X on the metabolic cost in healthy elderly, older than 65 years. It is important to first focus on healthy elderly that experience normal age-related biological degeneration before exoskeleton assistance can be applied in elderly with pathologies. Although it seems evident that WALL-X will result in a reduction in metabolic cost in healthy elderly, alterations in the walking pattern because of degeneration that comes with age can have an influence on the human-exoskeleton interaction. The application of exoskeletons in healthy elderly is a first step in our plans to implement exoskeletons in the rehabilitation and daily life of COPD patients and other populations with reduced exercise tolerance. We hypothesized that elderly could walk with the exoskeleton and that they could reduce the metabolic cost of walking as Norris et al. (Norris, Granata, et al., 2007) already showed that elderly are able to walk with an exoskeleton. Basic kinematics and perception measures were taken to evaluate the effect on balance and to study potential limitations in the applications of exoskeletons in elderly and on the long term in subjects with reduced exercise tolerance.

?	Can an exoskeleton be used to reduce the metabolic cost of walking in healthy elderly (aged 65 and over)?

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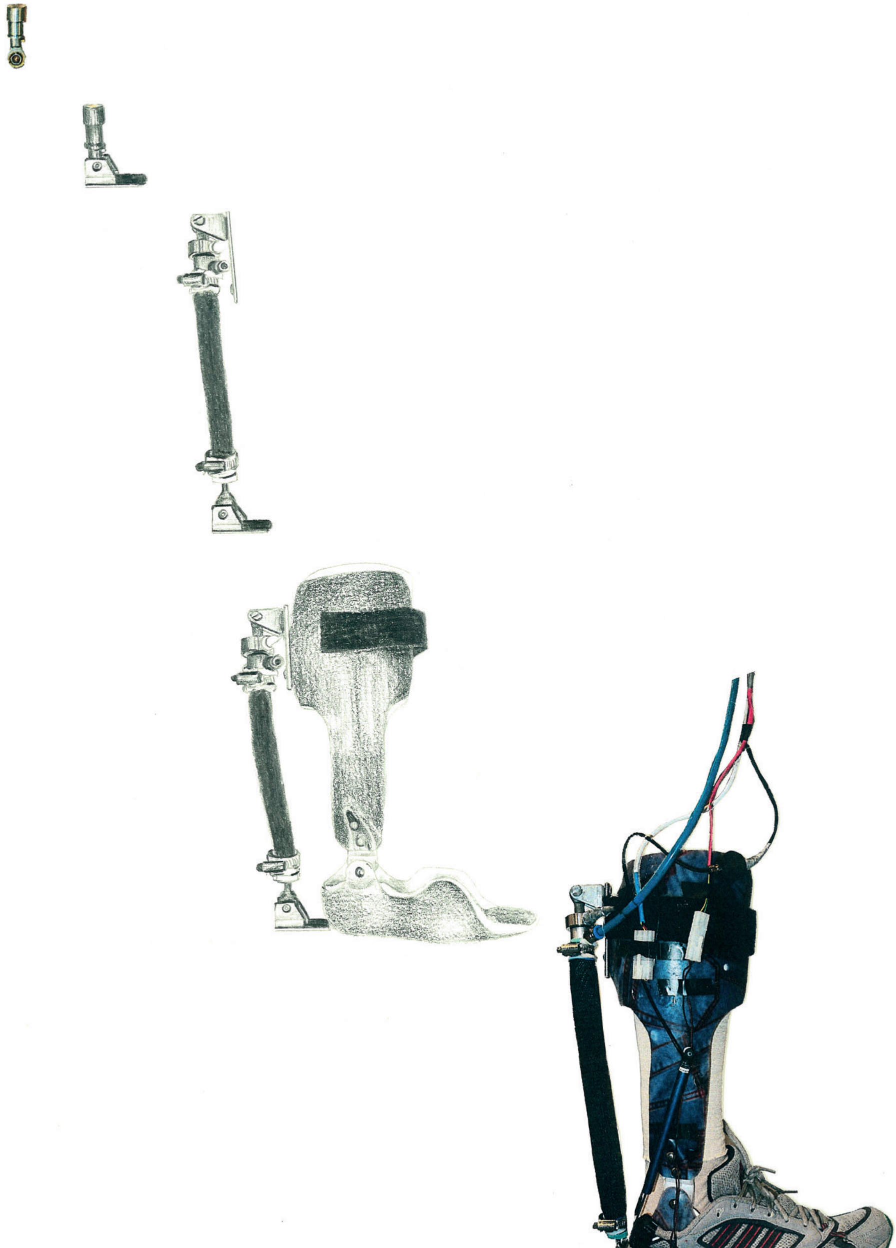
CHAPTER 1



DESIGN AND CONTROL OF WALL-X

1. Introduction

In all our studies we used a Wearable Assistive Lower Leg eXoskeleton, referred to as WALL-X. WALL-X was developed by Malcolm et al. in 2007 (Malcolm, Segers, et al., 2009), based on the design of Ferris et al. (Ferris, Czerniecki, et al., 2005). Ir. Davy Spiessens played a significant role in the development of the exoskeleton from the early beginning until the most recent optimizations and technically assisted in all experimental data collections that happened with the exoskeleton.

WALL-X is a simple ankle-foot exoskeleton that can be worn by healthy subjects during walking. Both hardware and software were revised and improved over the course of the consecutive studies of this thesis and a description of the improvements show some of the work, apart from extensive pilot testing and maintaining the exoskeleton, that is done during the period of this thesis and which cannot be found in the further mentioned chapters. This detailed description of WALL-X also emphasizes the simplicity of the exoskeleton, the simple control mechanism and the relatively low cost.

The improvements in WALL-X over the course of this thesis allowed to perform more advanced research. There was a continuous interaction between exoskeleton improvements and the research that could be done with the exoskeleton. In the beginning, the exoskeleton was used for gait transition studies in which the exoskeleton had to perform continuously for 30s (Malcolm, Fiers, et al., 2009; Malcolm, Segers, et al., 2009). Later on we wanted to do metabolic cost measurements, which implied that the exoskeleton had to perform continuously for at least 4 min (Malcolm et al., 2013). Further studies included continuous efforts of 24 min (*chapter* 2) or even subsequent 3 min intervals of 3 min that together summed to 30 min or more (*chapter* 5). In our most recent experiments the challenge was to develop a control mechanism that allowed to control average exoskeleton ankle joint mechanical power during walking (*chapter 3*). For each experiment, the exoskeleton had to meet certain criteria and the exoskeleton characteristics defined the limits for possible research protocols.

At the moment our exoskeleton is a stable device that allows to conduct experiments with more experimental conditions than any other lab in the world, which is due to the huge amount of time that is invested in extensive pilot testing and optimizing the exoskeleton hardware. More than 100 different subjects have been walking with our exoskeleton during the time period of this thesis. In June 2014 a hardware demonstration was done in Zurich on the Dynamic Walking conference (a video of this demonstration can be found on https://www.youtube.com/watch?v=jAUxuX2h7_U). Conference attendants were able to walk with the exoskeleton and experience the assistive effect of the exoskeleton themselves (Fig. 1). In preparation of the demonstration we changed our set-up so that all the hardware fits into a transportable case. At the moment, it only takes 30 minutes to set-up our exoskeleton on a new location and it takes about 5 minutes to prepare a subjects to walk with the device. This emphasizes the durability and the performance of our current exoskeleton.

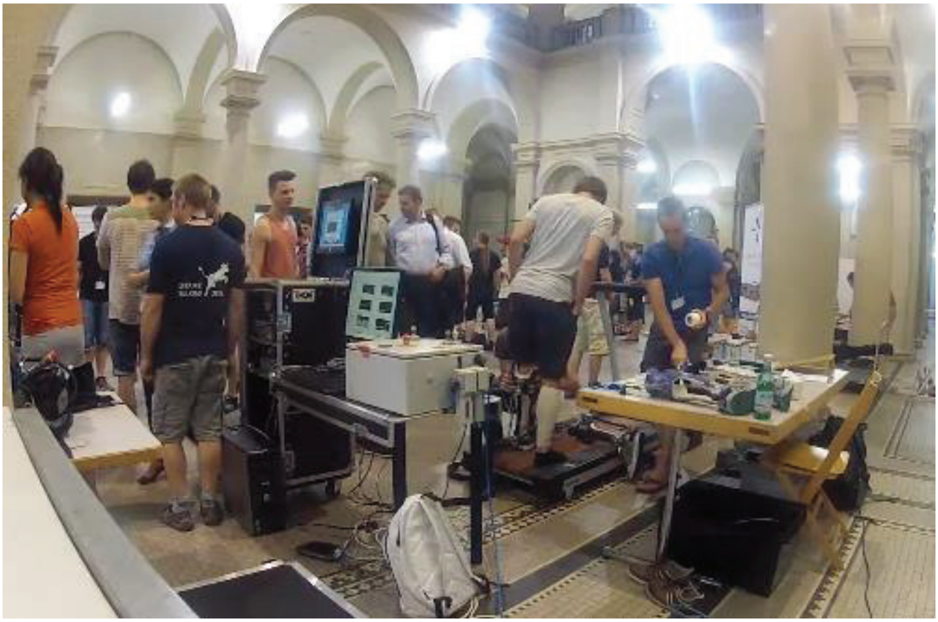


Fig. 1
Set-up of the exoskeleton and the hardware to control the exoskeleton on the Dynamic Walking conference in Zurich in 2014. 15 attendants of the conference walked with our exoskeleton to experience the assistive effect during walking.

2. Exoskeleton torque, power and work

One of the most important measures that describes the exoskeleton assistance of WALL-X is the exoskeleton ankle joint mechanical power, which is the exoskeleton work performed per unit time (Winter, 1991). Therefore, some more information on the specific parameters to calculate this exoskeleton power and the terminology that is used in each chapter is given.

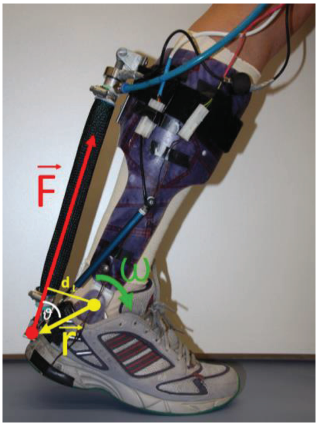


Fig. 2

WALL-X during walking with the force vector of the pneumatic muscles $\vec{F}$, the position vector $\vec{r}$, the angle between the position and the force vectors θ, the perpendicular distance between the ankle joint and the force vector $d_{\perp}$ and the joint angular velocity ω.

In WALL-X and other comparable exoskeletons, the tension force of the pneumatic muscle F (Fig. 2) causes an exoskeleton torque τ. Dis torque τ is the product of a force acting at a distance about an axis of rotation and causes an angular acceleration about that axis (Winter, 1991). It is calculated as the cross product of the force vector and the position vector:

$$\vec{\tau} = \vec{F} \; x \; \vec{r}.$$

This torque could be defined as the tendency of a force to rotate an object about an axis (Tipler and Mosca, 2009). The magnitude of the exoskeleton torque is calculated as follows:

$$\tau = F \; r \; sin\theta,$$

$$\tau = F \; d_{\perp},$$

with the magnitude of the force F measured with a load cell (Fig. 3A). The perpendicular distance between the pneumatic muscle and the joint angle, referred to as $d_{\perp}$, can be measured with motion capture (Fig. 3B). By multiplying both, the exoskeleton torque can be calculated for the duration of a stride (Fig. 3C). This exoskeleton torque profile is used in most chapters to determine the pneumatic muscle actuation onset and actuation ending (Fig. 3C).

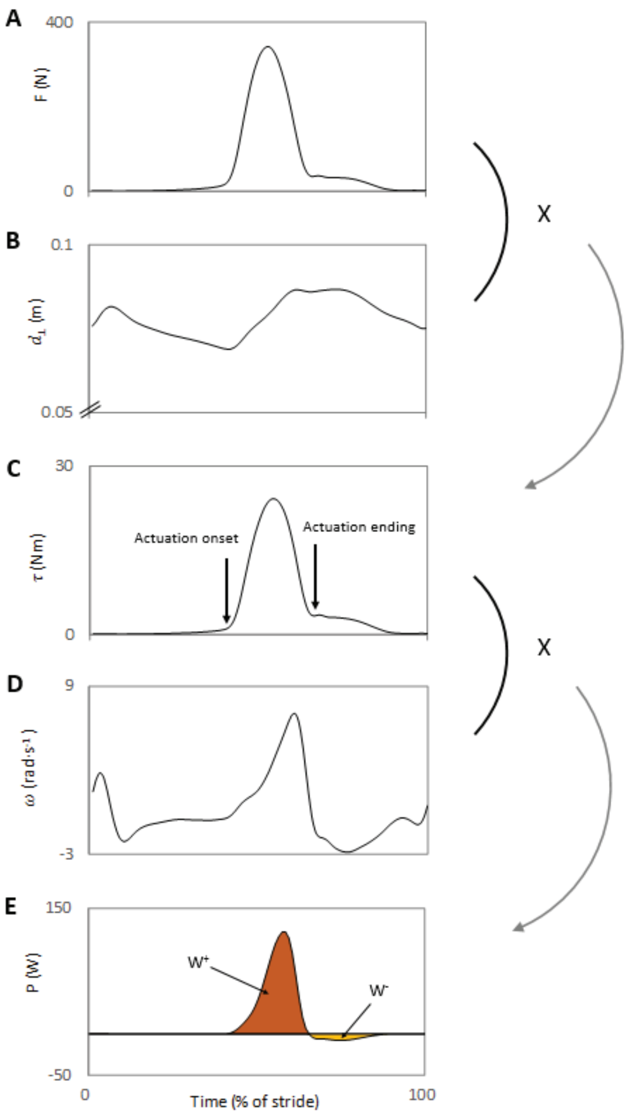


Fig. 3

Data of exoskeleton kinetics during walking with WALL-X, based on the results of *chapter* 3. Results are stride averages from heel contact to heel contact for the pneumatic muscle force in Newton (N), the moment arm of the pneumatic muscle force in meter (m), the torque of the exoskeleton in Newtonmeter (Nm), the ankle joint angular velocity in radians per second (rad·s^{-1}) and the exoskeleton ankle joint mechanical power in Watts (W) with positive (W^+) and negative (W^-) amounts of work.

The exoskeleton ankle joint mechanical power P is calculated as the product of the exoskeleton torque and the ankle joint angular velocity (Winter, 1991). The magnitude of the exoskeleton ankle joint mechanical power can be calculated (Fig. 3E) by multiplying the exoskeleton torque (Fig. 3C) with the ankle joint angular velocity (Fig. 3D), which is also measured with motion capture:

$$P = \tau\,\omega.$$

This exoskeleton ankle joint mechanical power allows to calculate the amount of positive (W^+) and negative work (W^-) that is delivered by the exoskeleton during a step (Fig. 3E). The amount of work that is delivered by the exoskeleton over a specific time interval is calculated by numerical integration of the exoskeleton ankle joint mechanical power over that time interval. This time interval can be a positive burst to calculate the positive work (W^+), a negative work burst to calculate the negative work (W^-), or the entire stride to calculate the net work per stride ($W^+ + W^-$). For a given time interval the amount of work that is performed during that time interval (ΔW) is calculated as follows:

$$\Delta W = \int P(t)\,dt.$$

This could also be calculated by multiplying the average exoskeleton ankle joint mechanical power over this time interval with the duration of the time interval:

$$\Delta W = P_{avg}\Delta t.$$

If this is calculated over the entire stride, the work is often expressed as a specific amount of work (in Joules) per stride. However, stride time should be taken into account when comparing different exoskeleton conditions. In example, when the net work in two different conditions is the same but the stride time is twice as long in the first condition compared to the second condition, the work per stride will be the same. As a result, work parameters should always be expressed in relation to a time parameter, e.g. work per stride. Therefore, the average exoskeleton ankle joint mechanical power over the stride P_{avg} is calculated as an alternative parameter to evaluate the work:

$$P_{avg} = \frac{\Delta W}{\Delta t},$$

with Δt as the duration of the stride and ΔW the net work over the stride. This average power takes into account differences in stride time across subjects and could be calculated for both positive and negative power as follows:

$$P^+_{avg} = \frac{\Delta W^+}{\Delta t},$$

$$P^-_{avg} = \frac{\Delta W^-}{\Delta t},$$

with Δt as the duration of the stride, ΔW^+ as the positive work over the stride and ΔW^- as the negative work over the stride. Another reason to calculate average power in Watts (or Joules per second) instead of work in Joules is that we are mainly interested in the relation with the metabolic cost of walking. This metabolic cost of walking is often expressed as metabolic power in Watts with the formula of Brockway

(Brockway, 1987) and could thereby easily be related to the average exoskeleton power. Some studies refer to this average power as net work per unit time (Malcolm et al., 2015) or stride-averaged work rate (Caputo and Collins, 2014a), also expressed in Watts. Throughout the different chapters of this thesis, terminology sometimes differs due to differences between journals and/or comments of reviewers. In general, timing refers to the actuation onset timing and exoskeleton power refers to the average (positive) exoskeleton ankle joint mechanical power. The following table (Table 1) gives an overview of the used terminology and units for exoskeleton assistance parameters in each chapter based on the terminology that is explained in this paragraph.

Table 1: terminology of exoskeleton actuation parameters

TERM	EXPLANATION	UNIT	CHAPTER 1	CHAPTER 2	CHAPTER 3	CHAPTER 4	CHAPTER 5	CHAPTER 6
F	Tension force of the pneumatic muscle	Newton (N)	Pneumatic muscle force (N)	Pneumatic muscle force (N)	Pneumatic muscle force (N)	/	Pneumatic muscle force (N)	Pneumatic muscle force (N)
$\boldsymbol{d}_{\perp}$	Moment arm of the pneumatic muscle force	Meter (m)	Moment arm of the pneumatic muscle force (m)	Moment arm of the pneumatic muscle (m)	Moment arm of the pneumatic muscle force (m)	/	Moment arm of the pneumatic muscle (m)	Moment arm of the pneumatic muscle force (m)
$\boldsymbol{\tau}$	Exoskeleton torque	Newtonmeter (Nm)	Exoskeleton torque (Nm)	Torque of the pneumatic muscle (Nm)	Exoskeleton torque (Nm)	/	Pneumatic muscle torque (Nm)	Exoskeleton torque (Nm)
	Start and ending of exoskeleton actuation	% of stride	Actuation onset and ending	Actuation onset and offset	Actuation onset and ending	Actuation onset and offset	Start of pneumatic muscle actuation and turning off	Actuation onset and ending
$\boldsymbol{\omega}$	Angular velocity around the exoskeleton ankle joint	Radians per second ($rad \cdot s^{-1}$) or degrees per second ($° \cdot s^{-1}$)	Ankle joint angular velocity ($rad \cdot s^{-1}$)	Ankle joint angular velocity ($rad \cdot s^{-1}$)	Ankle joint angular velocity ($rad \cdot s^{-1}$)	Ankle angular velocity ($° \cdot s^{-1}$).	Ankle joint angular velocity ($rad \cdot s^{-1}$)	Ankle joint angular velocity ($° \cdot s^{-1}$)
P	Exoskeleton ankle joint mechanical power	Watts or Joule per second ($W = J \cdot s^{-1}$)	Exoskeleton ankle joint mechanical power ($W \cdot kg^{-1}$)	Exoskeleton mechanical power	Exoskeleton ankle power	/	Exoskeleton mechanical power	Exoskeleton ankle joint mechanical power
$\boldsymbol{P_{avg}}$	Average exoskeleton ankle joint mechanical power per stride (= net	Watts or Joule per second ($W = J \cdot s^{-1}$)	Average exoskeleton ankle joint	/	/	/	Net exoskeleton mechanical power per stride	/

	work divided by stride time)		mechanical power					
P^{+}_{avg}	Average positive exoskeleton ankle joint mechanical power per stride (= positive work divided by stride time)	Watts or Joule per second (W = $J{\cdot}s^{-1}$)	Average positive exoskeleton ankle joint mechanical power	/	Average positive exoskeleton ankle power (summed for both legs)	/	/	Average positive mechanical exoskeleton power

3. Actuation control of WALL-X

Our goal is to study the interaction between exoskeleton assistance and the metabolic cost of walking. Therefore, we are not concerned about making an autonomous exoskeleton where all the hardware and power sources are carried by the user and thus allows to use it outdoor or in real life situations. Scientific research should also focus on fundamental research questions and show how exoskeletons could assist locomotion. Companies can than focus on building these devices, make them lightweight and autonomous and bring them to consumers. We used an exoskeleton that can be used on treadmill and is tethered to a stationary power source, compressed air supply and off-board hardware. WALL-X should therefore not be seen as a commercial device but as a stationary testbed (Fig. 4) that allows to manipulate exoskeleton actuation parameters in an easy way. By doing so we can study the metabolic response to changes in exoskeleton actuation, also during walking in challenging environments (uphill walking and walking while carrying weights) or in challenging populations (elderly or patients with COPD). While our exoskeleton can be used in a stationary setting, e.g. during rehabilitation, our optimal actuation parameters and our findings on the human-exoskeleton interaction during walking in challenging environments or in challenging populations can also be implemented in autonomous and practical devices.

Fig. 4
Example of the stationary exoskeleton testbed at the Ghent University. The subject is wearing the exoskeleton, which is connected to off-board hardware that controls the pneumatic muscle actuation by compressed air supply to the pneumatic muscles.

WALL-X consists of a thermoplastic orthosis with pneumatic muscles attached at the posterior side that can assist plantar flexion during the push-off (Fig. 5). All orthoses in our studies were made by Aqtor! (previously Technische Orthopedie België) and converted to pneumatic exoskeletons by the lab of Movement Sciences of the Ghent University. Every study used one pair of exoskeletons, designed to fit the mean anthropometry of the targeted populations and one shoe size that fitted more or less for all subjects.

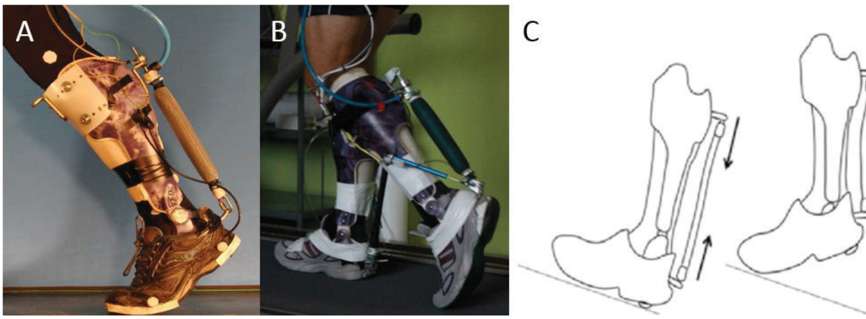


Fig. 5
WALL-X (A and B), consisting of a thermoplastic orthosis with pneumatic muscles attached at the posterior side. These pneumatic muscles, when filled with compressed air, can shorten and can assist plantar flexion of the ankle during walking. The exoskeleton itself is lightweight (app. 0.7 kg) but can only be used on a treadmill due to off-board hardware, power and air supply. Figure A shows the exoskeleton as in *chapter* 2, 4 and 5 (with a tension load cell mounting that did not yet measured pneumatic muscle force with sufficient precision). Figure B shows the exoskeleton of *chapter* 3 and 6 with the improved tension load cell and the additional linear displacement sensor (blue). Figure C shows the effect of the pneumatic muscle contraction on ankle joint angle (during uphill walking), causing plantar flexion when assisted during the push-off in walking.

The control of the pneumatic muscles actuation works with custom software (Labview, National Instruments, TX, USA) based on a simple and straightforward algorithm (Fig. 6). The pneumatic muscles are connected with a pneumatic valve station (Festo, Esslingen, Germany) that regulate air pressure and air supply to the pneumatic muscles based on software that allows to set the preferred onset and ending of actuation as a percentage of the stride time (e.g. 40 and 60% of the gait cycle respectively). Based on footswitches in the heel, stride time is determined (e.g. 1.05s) and the software calculates the onset (0.40 x 1.05s) and ending of actuation (0.60 x 1.05s) of the pneumatic muscles, using the imposed percentages of the stride time. Following the next heel contact, the software steers out an onset and ending signal to the control box (after respectively 0.42 and 0.63s), corrected for the delay between the time of steering out the signal and the actual start of pneumatic muscle actuation, that opens and closes pneumatic pressure valves and allows the pneumatic muscles to contract and relax.

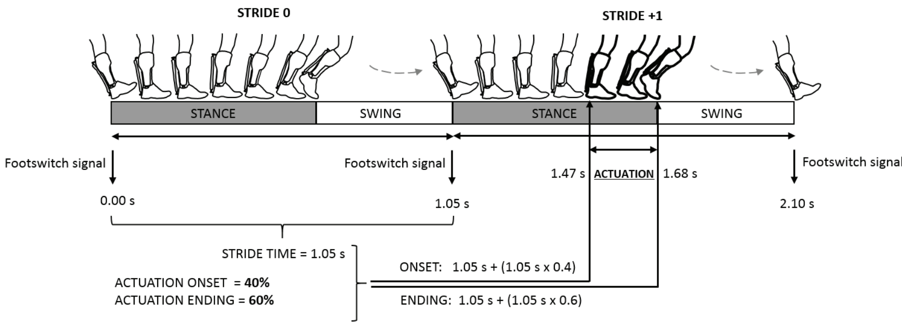


Fig. 6

Schematic representation of the simple control algorithm for the actuation timing of the pneumatic muscles. Based on the stride time of a previous stride (STRIDE 0), actuation onset and ending for the following stride (STRIDE +1) is determined using a fixed actuation onset and ending based on the stride time and the preferred settings. The exact same happens for the opposite leg and for all the following strides.

In *chapter* 2, 4 and 5, the control algorithm only used timing of heel strike to determine actuation onset and ending of the pneumatic muscles, using an imposed actuation onset, actuation ending and a fixed air pressure as settings (Fig. 7A). Air pressure was set at 3 Bar, based on pilot tests. A load cell and joint kinematic measurements allowed to calculate exoskeleton ankle joint mechanical power assistance after the experiment, using ankle joint angular velocity, the moment arm of the pneumatic muscle force and the pneumatic muscle force.

In *chapter* 3 and 6, a linear displacement sensor (SLS130, Penny&Giles, Christchurch, United Kingdom) was added to the exoskeleton, which allowed to measure ankle joint angular velocity and moment arm of the pneumatic muscle force in real-time during walking. Thereby, it was possible to calculate exoskeleton mechanical power assistance in real-time during walking with the exoskeleton (Fig. 5B and Fig. 7B). This improvement allowed to adjust air pressure in order to reach a certain amount of average exoskeleton power assistance as increased air pressure results in increased pneumatic muscle forces (Fig. 9). We developed an iterative learning controller that allowed to automatically adjust air pressure of the pneumatic muscles in order to reach a desired amount of average exoskeleton power. The software calculated the difference between the desired amount of average exoskeleton power and the actual amount of average exoskeleton power and slowly adjusted air pressure in order to reach the desired amount of average exoskeleton power. The settings that were used as input variables for the exoskeleton control during these experiments were the actuation onset, the actuation ending and the desired amount of average exoskeleton power over a stride. This approach allowed to do more controlled experiments as it allowed to apply different actuation timing conditions, while keeping average exoskeleton power at a fixed level or to apply different amounts of average exoskeleton power, while keeping actuation timing at a fixed level. In *chapter* 6, the linear displacement sensor was indeed used to calculate exoskeleton mechanical power during the experiment but the iterative learning controller was not used to adjust air pressure. Air pressure was kept at 2 Bar for all conditions and all subjects in *chapter* 6 in order to deliver relatively low amounts of exoskeleton power for the elderly.

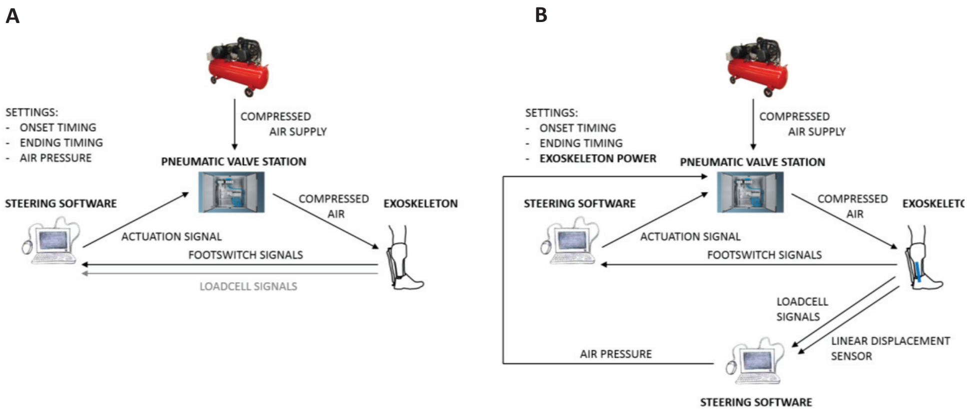


Fig. 7

The exoskeleton set-up used in *chapter 2*, 4 and 5 (A) allowed to impose an actuation timing and a fixed air pressure for the pneumatic muscles. This resulted in an exoskeleton actuation with a certain amount of average exoskeleton power, which could be measured after the experiment. The only settings that could be set were the onset timing, the ending timing and the air pressure of the pneumatic muscles. The exoskeleton set-up in *chapter 3* and 6 (B) allowed to measure average exoskeleton power in real-time during walking by means of a linear displacement sensor. With an iterative learning controller the air pressure of the pneumatic muscles could be adjusted in order to reach a desired amount of average exoskeleton power during walking. The settings that could be set where the onset timing, the ending time and the desired average amount of exoskeleton power.

4. Other improvements to WALL-X

Also some specific improvements to the ankle-foot orthosis, the footswitches and the pneumatic muscles were done over the time course of the thesis. The first exoskeleton, which was used for transition studies (Malcolm, Fiers, et al., 2009; Malcolm, Segers, et al., 2009), had a distal connection for the pneumatic muscles at the heel cup. As this caused discomfort at the heel during longer walking in several subjects, it was changed to a metal plate at the plantar side of the foot and levelled with cork (Fig. 8A and Fig. 8B). This renewed construction was used in all studies that are presented in this thesis. In *chapter* 2 and 4, rather large switches (IP67, Herga Electric, Suffolk, United Kingdom) were stripped and placed in the sole of a shoe, covered with a thin metal cover (Fig. 8C and Fig. 8E) to detect heel contact during walking. As this construction was very sensitive to wear and tear, it was hard to continue longer walking efforts without interruptions. Therefore, the footswitches were changed into smaller switches (Multimec 5E/5G , Mec, Ballerup, Denmark) with a plastic cover of 2mm in the sole of the shoe, which appeared to be less sensitive to wear and tear (Fig. 8D). These switches were used in *chapter* 3, 5 and 6.

In order to allow longer periods of continuous walking in *chapter* 4, additional air filters (Fig. 8F) were added between the pneumatic valve station and the pneumatic muscles to prevent dust from the pneumatic muscles to enter the pneumatic valve station. Unfortunately they had an unanticipated negative effect on timing of pneumatic muscle actuation and were again removed for *chapter* 5 and 6. In order to prevent water to enter the pneumatic valve station, additional filters were placed between the compressed air supply and the pneumatic valve station in *chapter* 5 and 6, which did not influence pneumatic muscle actuation timing.

A load cell allowed to measure the tension force of the pneumatic muscles during walking. In *chapter* 2, 4 and 5, a clamped pressure load cell was used (A.L. Design, Buffalo, NY, USA) but frequent use lead to non-linear behaviour of the load cell and resulted in a presumably overestimated pneumatic muscle force in *chapter* 2 and unusable load cell data in *chapter* 4 and 5 (Fig. 5A and Fig. 8G). Therefore, an improved pneumatic muscle connection was designed by the Department of Materials Science and Engineering of the Ghent University, which allowed us to use a tension load cell (210 Series, Richmond Industries Ltd., Reading, UK) and resulted in accurate and reliable load cell data in *chapter* 3 and 6 (Fig. 5B and Fig. 8H). These new load cells were extensively tested during bench-top testing and during human walking and showed accurate and reliable results.

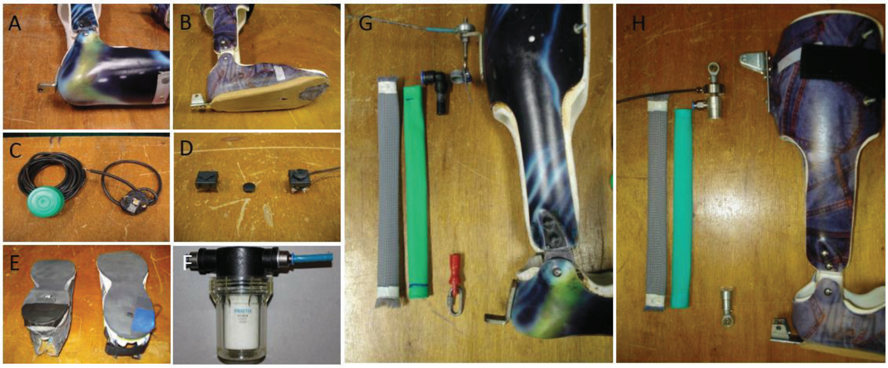


Fig. 8

Examples of technical improvements over the different studies: improvements of the pneumatic muscle connection to the heel of the exoskeleton (A and B), improvements of the footswitches in order to allow continued walking (C, D and E), additional air filters to remove water from the compressed air supply to allow longer usability of the pneumatic muscles (F) and improvements in the load cell and the pneumatic muscle connection to the exoskeleton (G and H).

The pneumatic muscles that were used in our lab (Fig. 5A, Fig. 8G and Fig. 8H) consist of a standard cycling tube and an expandable braided sleeve with a diameter of 30 mm and a length around 0.28 m, dependent on the used orthosis. Pneumatic muscle length needs to be long enough to prevent passive pneumatic muscle forces during dorsiflexion but short enough to allow high peak forces. In our studies, we adjusted pneumatic muscle insertion to allow 15° of dorsiflexion, which is enough to allow normal walking (Winter, 1991).The force-length relationships of the pneumatic muscles (Fig. 9) showed a linear relationship during bench-top testing, which is in accordance with published characteristics of pneumatic muscles (Gordon et al., 2006).

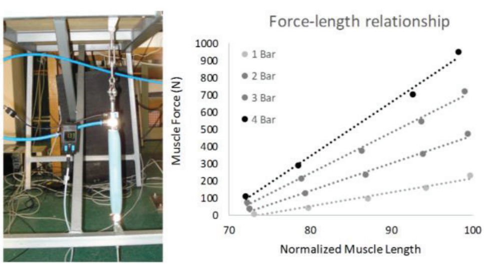


Fig. 9

Force-length relationship of the pneumatic muscles, determined during a static set-up of the pneumatic muscles (left). Longer muscle lengths lead to higher forces. The force-length relationship is depending on the air pressure, leading to higher pneumatic muscle forces when air pressure is higher.

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CHAPTER 2

ADAPTATION TO WALKING WITH AN EXOSKELETON THAT ASSISTS ANKLE EXTENSION

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Adaptation to walking with an exoskeleton that assists ankle extension

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1. Abstract

The goal of this study was to investigate adaptation to walking with bilateral ankle-foot exoskeletons with kinematic control that assisted ankle extension during push-off. We hypothesized that subjects would show a neuromotor and metabolic adaptation during a 24 min walking trial with a powered exoskeleton. Nine female subjects walked on a treadmill at 1.36 ± 0.04 ms^{-1} for 24 min with a powered exoskeleton and for 4 min with an unpowered exoskeleton. Subjects showed a metabolic adaptation after 18.5 ± 5.0 min, followed by an adapted period. Metabolic cost, electromyography and kinematics were compared between the unpowered condition, the beginning of the adaptation and the adapted period. In the beginning of the adaptation (4 min), a reduction in metabolic cost of 9 % was found compared to the unpowered condition. This reduction was accompanied by reduced muscular activity in the plantar flexor muscles, as the powered exoskeleton delivered part of the necessary ankle extension moment. During the adaptation this metabolic reduction further increased to 16 %, notwithstanding a constant exoskeleton assistance. This increased reduction is the result of a neuromotor adaptation in which subjects adapt to walking with the exoskeleton, thereby reducing muscular activity in all leg muscles. Because of the fast adaptation and the significant reductions in metabolic cost we want to highlight the potential of an ankle-foot exoskeleton with kinematic control that assists ankle extension during push-off.

2. Introduction

Because of the frequent use of walking in daily life and the walking recommendations to increase quality of life [1], a robotic exoskeleton that assists walking has the potential to become an assistive device in disabled people or a performance-increasing tool when walking endurance is challenged. Although the technical aspect of exoskeletons and powered prostheses is developing rapidly, little attention is paid to how people interact with exoskeletons [2–4] whilst this is an unavoidable step in the development of an exoskeleton for daily use. In this context, researchers attribute great potential to pneumatically powered ankle-foot exoskeletons [2,3]. These exoskeletons consist of a shell around the lower leg and can be powered by compliant pneumatic actuators, operating as artificial muscles. Similar to human skeletal muscles these actuators can contract, thereby assisting ankle extension when origin and insertion are located at the rear side of the exoskeleton.

The metabolic cost of walking with a powered exoskeleton can be lower than that of walking with an unpowered exoskeleton. Sawicki and Ferris [5–7] found reductions up to 13 % for a bilateral exoskeleton with a control algorithm based on m. soleus electromyography (EMG) after an adaptation period up to 90 min. A drawback of these EMG-based control algorithms is that they need a relatively long adaptation period. This could be explained by the continuous interaction between the control algorithm (based on EMG), and the neuromotor adaptation to walking with the exoskeleton that will alter these EMG signals. Indeed, Norris et al. [8] reported an increased metabolic efficiency of 15 % for powered versus unpowered walking after only 7 min of adaptation, using a more simple kinematic control method that assisted ankle extension. Malcolm et al. [9] showed that the kinematic control method can be done based on timing of stance phase with metabolic reductions up to 17 %. One study compared a kinematic control method with an EMG control method [4] and favoured EMG but their kinematic control method was activated too early during stance according to the findings of Malcolm et al. [9].

There seems a discrepancy between the two control methods and although most attention is faced toward EMG control the theoretical modelling of Kuo [10], which is supported by findings of Malcolm et al. [9], highlights the potential of an exoskeleton with kinematic control and optimal actuation timing. However, adaptation to an exoskeleton with kinematic control is not yet investigated in detail. Therefore, we aimed to investigate the adaptation to simple bilateral ankle-foot exoskeletons with ankle extension assistance during a 24 min walking trial [11] in terms of the metabolic adaptation, the neuromotor adaptation, and the interaction between both. We hypothesized that subjects will show a neuromotor adaptation in their kinematics and muscle recruitment, and as such have accommodated metabolically within this adaptation period.

3. Methods

Subjects

Nine healthy female subjects (age 20.8 ± 0.7 years; body mass 61.4 ± 4.9 kg; stature 167.8 ± 5.0 cm; means ± s.d.) participated in the experiment. They had no previous experience with exoskeleton walking. All participants signed an informed consent, approved by the ethical committee of the Ghent University hospital.

Exoskeletons

The bilateral exoskeletons were constructed similar to other ankle-foot exoskeletons [4–7,11-13]. They had a weight of 0.76 kg each and fitted in running shoes where footswitches were built in (IP67, Herga Electric, Suffolk, UK). Based on footswitch signals, a computer program (Labview, National Instruments, Austin, TX, USA) triggered onset and offset of the pneumatic muscles [14] actuation by means of a feedforward algorithm. Actuation timing of the pneumatic muscles was set at 43 % and deactivation timing at 63 % of stride [9].

Protocol

Each subject completed a powered and unpowered condition in randomized order. Treadmill was set at a constant dimensionless speed of 0.47 (e.g. 1.36 ms^{-1} for a leg length of 0.90 m)[15]. In the powered condition, subjects walked for 24 min as this was the adaptation period reported by Gordon and Ferris [11] for a unilateral exoskeleton with EMG control. In the unpowered condition subjects walked for 4 min as pilot testing showed that this was sufficient to metabolically adapt to unpowered walking.

Data collection

Lower limb kinematics were analyzed by 20 reflective markers on the right lower limb, recorded with 11 infrared camera's (200 Hz; Pro Reflex, Qualisys AB, Gothenburg, Sweden) and Qualisys software. EMG was measured (200Hz; Zerowire, Noraxon, Scottsdale, AZ, USA) with surface electrodes that were placed in accordance with SENIAM guidelines [16] on the m. soleus (bilateral), medial m. gastrocnemius (bilateral), m. tibialis anterior (bilateral), m. vastus lateralis (right leg) and m. rectus femoris (right leg). Because of difficulties with electrode placement beneath the exoskeleton, EMG measurements were successful in only 7 subjects. Spatiotemporal parameters were calculated based on recordings of a high speed camera (200 Hz; Bassler, Ahrensburg, Germany) with MaxTraq software (Innovasion Systems, Columbiaville, MI, USA). Tensile force of the right pneumatic muscle was measured with a loadcel (200 Hz; A.L. Design, Buffalo, NY, USA). Kinematics, EMG, high-speed video and loadcell data were synchronized and collected during 10 s every minute. O_2 consumption and CO_2 production were recorded during the entire experiment (Oxycon Pro, Jaeger GMBH, Höchberg, Germany).

Data processing

Metabolic data were calculated with the formula of Brockway [17]. Resting metabolic cost was subtracted from metabolic data in the walking conditions and normalized by bodyweight to calculate net metabolic cost. During the powered condition the lowest 2-min-average was considered the minimum. The average of the individual quadratic regressions (Fig. 1) of the metabolic cost every 30 s during the powered condition illustrates the adaptation (the time to reach the minimum), followed by the adapted period once the minimum was reached.

Reflective marker data were filtered with a 4th order Butterworth low-pass filter (cutoff frequency 6 Hz) and sagittal lower limb kinematics were calculated in Visual 3D (C-Motion, MD, USA). Ankle and knee angles for the right leg were time-normalized from right heel contact to right heel contact (Fig. 2). Moment arm of the pneumatic muscle was calculated as the distance between the ankle joint and the pneumatic muscle of the right leg, which varied over the gait cycle. Torque of the right pneumatic muscle was calculated by multiplying pneumatic muscle force with moment arm. We calculated mechanical power of the exoskeleton by multiplying torque with ankle joint angular velocity and divided this by body mass. Exoskeleton mechanical power was also time-normalized from right heel contact to right heel contact (Fig. 2).

EMG data were rectified and moving root mean square (RMS) was taken to create linear envelopes which were averaged and time-normalized from right heel contact to right heel contact (Fig. 3). The mean RMS per stride was a measure for the activity of the muscles per distance as the absolute speed in the powered and unpowered condition was similar between subjects (1.36 ± 0.04 ms^{-1}). To express changes in EMG amplitude, average values of the linear envelop per stride were normalized to the average of the unpowered condition (Fig. 4).

Statistics

Repeated measures ANOVA was done with SPSS Statistics 20 (IBM, Armonk, NY, USA) with Tukey Honestly Significant Difference post hoc tests ($p \leq 0.05$) to compare metabolic data (2-min-averages), spatiotemporal parameters (stride time and stance time), kinematics (mechanical power, ankle and knee angle every 10% of stride and instant of peak values) and EMG (instant of peak activation and stride averages) between three conditions: the unpowered condition (metabolic cost of min 2-4 in combination with EMG and kinematic data of the fourth min of the unpowered condition), the beginning of the adaptation (metabolic cost of min 2-4 in combination with EMG and kinematic data of the fourth min of the powered condition) and the adapted period (metabolic cost of min 16.5-18.5 in combination with EMG and kinematic data of the 24th min of the powered condition).

4. Results

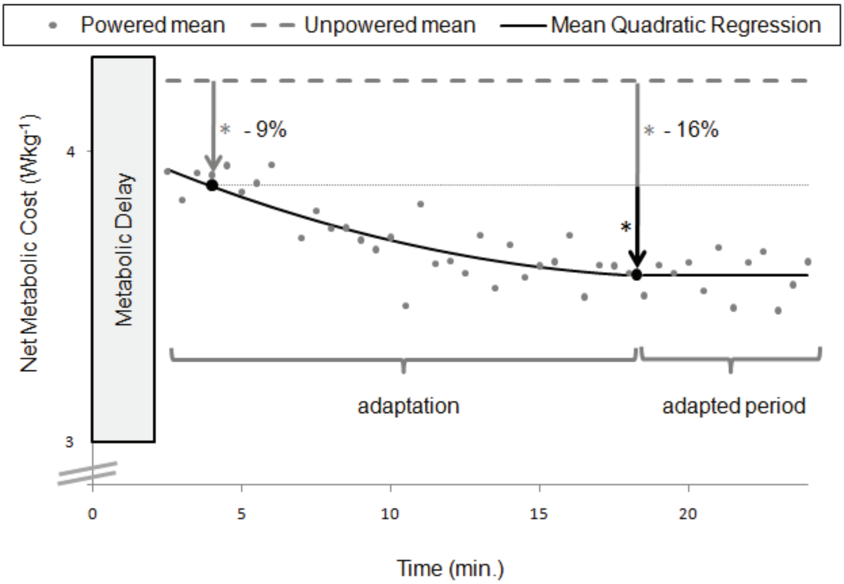

Fig. 1.
Net metabolic cost. Gray dots represent the mean net metabolic cost (30 s values) during the 24 min powered condition. Black solid line represents the mean quadratic regression over these 30 s values during the powered condition until the minimum is reached after 18.5 ± 5 min. The time to reach this minimum is called the adaptation, the time period after this minimum is called the adapted period. Gray dashed line represents the mean net metabolic cost of min 2-4 in the unpowered condition. Black dotted line represents the mean net metabolic cost of min 2-4 in the 24 min powered condition. Percentages represent reductions in net metabolic cost in the beginning of the adaptation (min 2-4) and in the adapted period (min 16.5-18.5) compared to the unpowered condition (min 2-4). Gray box represents the metabolic delay until steady state is achieved. *****= sign. different from the beginning of the adaptation (ANOVA, p ≤ 0.05); *= sign. different from the unpowered condition (ANOVA, p ≤ 0.05).

Subjects reached the minimum for metabolic cost in the 24 min powered condition after an average adaptation period of 18.5 ± 5.0 min. No subject showed the lowest metabolic cost in the last 2 min. After the adaptation a steady state in metabolic cost was achieved during the adapted period (Fig. 1), as net metabolic cost after the average adaptation time (3.56 ± 0.70 Wkg^{-1}; min 16.5-18.5) was not significantly different from the end of the powered condition (3.55 ± 0.68 Wkg^{-1}; min 22-24; p = 0.99). Therefore, net metabolic cost after the average adaptation time was taken as a reference for the entire adapted period (Fig. 1).

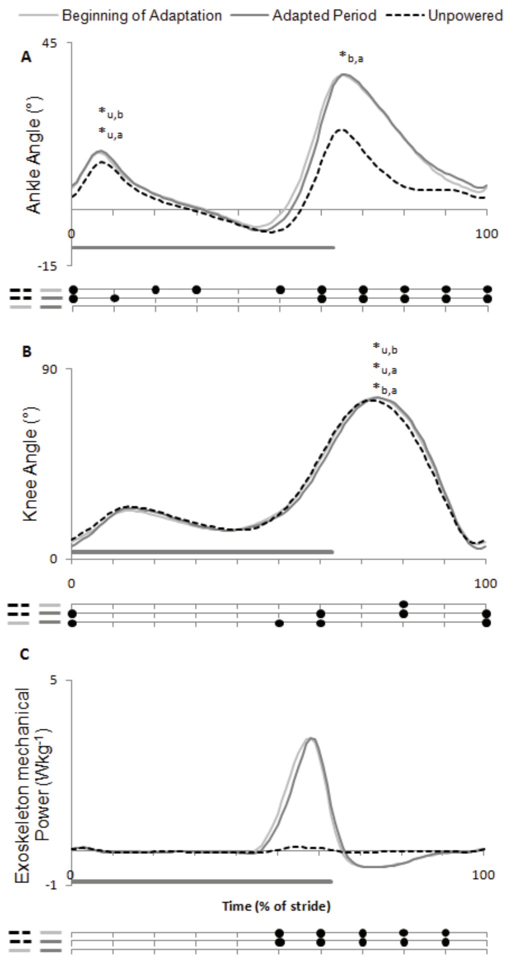

Fig. 2.
Ankle kinematics (A), knee kinematics (B) and exoskeleton mechanical power (C). Values are stride averages from heel contact (0 %) to heel contact (100 %) of the right lower limb for the beginning of the adaptation, the adapted period and the unpowered condition. Gray horizontal line represents stance phase (app. 64 % of stride). Statistical differences were computed every 10 % from 0 to 100 %. Dots at the bottom of each figure represent statistical differences (ANOVA, $p \leq 0.05$) every 10 % of stride between the 2 conditions which are printed at the left side: between unpowered and beginning of adaptation; unpowered and adapted period; beginning of adaptation and adapted period for respectively the first, second and third lines. Statistical differences in timing of peak activation (ANOVA, $p \leq 0.05$) are represented with * for differences between the beginning of the adaptation and the adapted period (*b,a); the unpowered condition and the beginning of the adaptation (*u,b) and the unpowered condition and the adapted period (*u,a).

Net metabolic cost in the unpowered condition (4.25 ± 0.59 Wkg^{-1}; min 2-4) was significantly higher than at the beginning of the adaptation (3.88 ± 0.73 Wkg^{-1}; min 2-4; $p \leq 0.05$) or than in the adapted period (3.56 ± 0.70 Wkg^{-1}; min 16.5-18.5; $p \leq 0.05$). During the adaptation the net metabolic cost significantly decreased from the beginning of the adaptation (3.88 ± 0.73 Wkg^{-1}; min 2-to-4) to the adapted period (3.56 ± 0.70 Wkg^{-1}; min 16.5-18.5; $p \leq 0.05$).

Stride time did not differ between the unpowered condition (1.06 ± 0.05 s), the beginning of the adaptation (1.06 ± 0.05 s) and the adapted period (1.08 ± 0.05 s). Differences in stance time were found between the adapted period (0.70 ± 0.03 s) and both the beginning of the adaptation (0.68 ± 0.03 s; $p \leq 0.05$) and the unpowered condition (0.68 ± 0.03 s; $p \leq 0.05$). Ankle kinematics showed similar patterns for the beginning of the adaptation and the adapted period with a significant difference in timing of second peak angle of 0.89 ± 0.60 % (Fig. 2A). More significant differences in the ankle were found between the unpowered condition and both the beginning of the adaptation and the adapted period with significant differences in timing of first peak angle of respectively 0.67 ± 0.50 % and 0.56 ± 0.53 % (Fig. 2A). Knee kinematics showed similar patterns for the unpowered and the powered conditions with some small statistical differences between the 3 conditions and a significant difference between the timing of the second peak angle of the unpowered condition (72.67 ± 1.32 %), the beginning of the adaptation (73.56 ± 1.33 %) and the adapted period (74.11 ± 1.17 %) (Fig. 2B). Exoskeleton mechanical power differed between the unpowered condition and both powered conditions when the exoskeleton was activated but did not significantly change during the powered condition (Fig. 2C).

Only the m. soleus showed a significant difference in timing of peak activation between the beginning of the adaptation (36.50 ± 4.81 %) and both the adapted period (41.36 ± 1.14 %; $p \leq 0.05$) and the unpowered condition (40.79 ± 2.14 %; $p \leq 0.05$) (Fig. 3). In the beginning of the adaptation the mean RMS in the m. soleus and m. gastrocnemius was significantly lower than in the unpowered condition while the m. tibialis anterior was significantly higher (Fig. 4). During the adaptation RMS per stride in all muscles decreased from the beginning of the adaptation to the adapted period, but only for the m. tibialis anterior, the m. soleus and the m. biceps femoris a significant difference was observed (Fig. 4). The mean RMS in the adapted period was significantly lower than in the unpowered condition for the m. soleus, the m. gastrocnemius and the m. biceps femoris (Fig. 4).

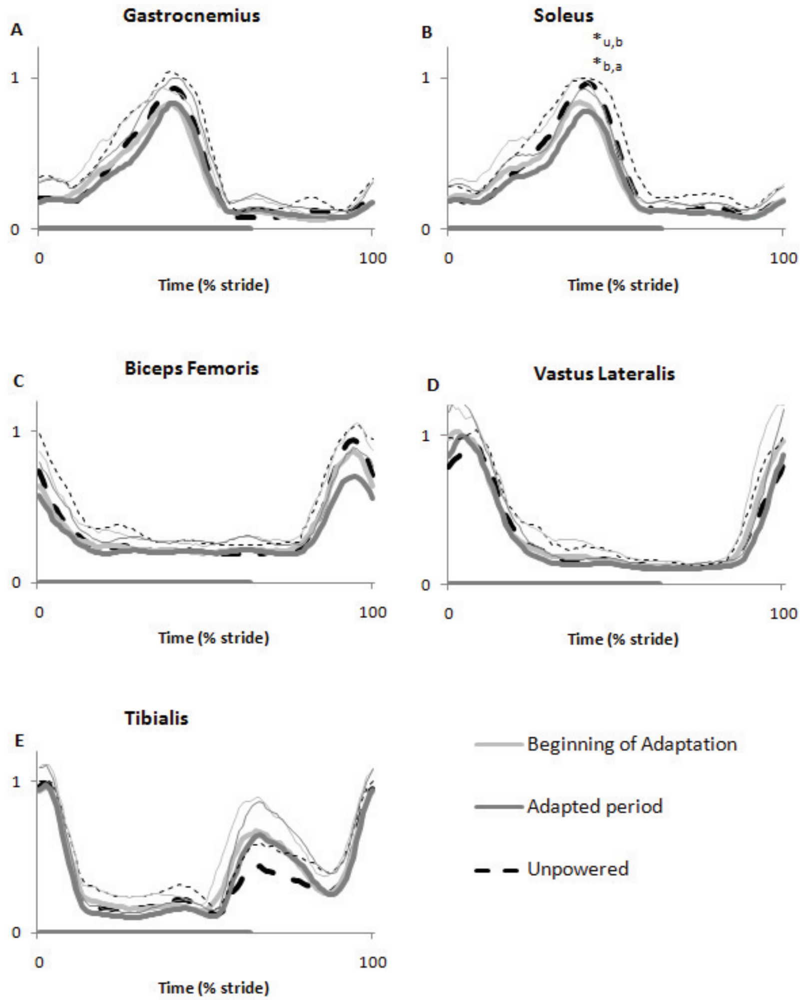

Fig. 3.
EMG RMS linear envelopes of 5 muscles. Curves are normalized to the peak value of the unpowered condition and averaged from heel contact (0 %) to heel contact (100 %). Mean linear envelopes (thick lines) + s.d. (thin lines) for m. gastrocnemius (A), m. soleus (B), m. biceps femoris (C), m. vastus lateralis (D) and m. tibialis anterior (E) are plotted for the beginning of the adaptation, the adapted period and the unpowered condition. Gray horizontal line represents stance phase (app. 64 % of stride). Statistical differences in timing of peak activation (ANOVA, $p \leq 0.05$) are represented with * for differences between the beginning of the adaptation and the adapted period (*b,a); the unpowered condition and the beginning of the adaptation (*u,b) and the unpowered condition and the adapted period (*u,a).

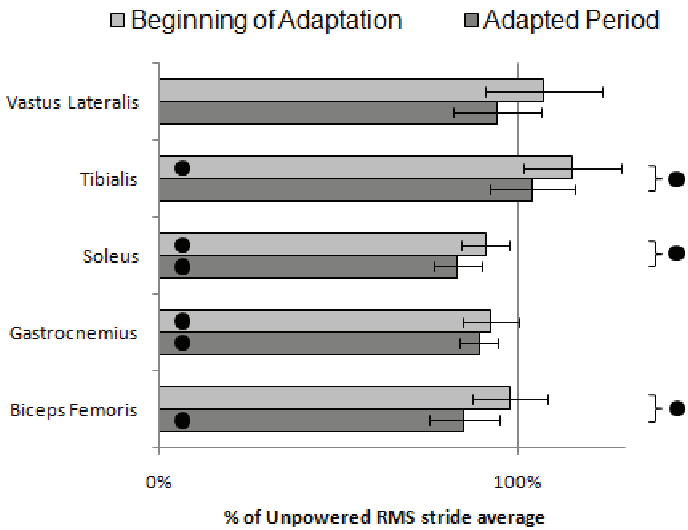

Fig. 4.

EMG RMS stride averages. Histograms for EMG RMS stride averages of 5 muscle groups, normalized to the mean RMS per stride for the unpowered condition. Error bars are ± s.d. Values above 100 indicate that a muscle has an increased activity compared to the unpowered condition. Values beneath 100 indicate that a muscle has a decreased activity compared to the unpowered condition. Dots in the bars represent significant differences from 100, which means significantly different from the unpowered condition (ANOVA, $p \leq 0.05$). Dots with parentheses at the right represents significant differences between both bars, which means significant differences between the beginning of the adaptation and the adapted period (ANOVA, $p \leq 0.05$).

5. Discussion

After only 4 min of powered exoskeleton walking, a significant reduction of 9% was found in metabolic cost as subjects used the power of the exoskeleton. The pneumatic muscles delivered additional power to the ankle joint during push-off. This caused increased plantar flexion and resulted in an increased antagonistic m. tibialis anterior activity in the beginning of the adaptation, probably in an attempt to restore normal joint angles before heel contact. Notwithstanding this perturbation of normal ankle kinematics and m. tibialis anterior EMG, the exoskeleton assistance resulted in an almost immediate reduction in metabolic cost of 9 % compared to unpowered walking. This indicates the ability of humans to use (part of) the additional power of the exoskeleton in the ankle [7], thereby reducing the necessary biological ankle extension moment. As a consequence of the lower moment that must be delivered by the subjects' own ankle musculature, a reduction in muscle activity of the m. soleus and the m. gastrocnemius was found. It seems logical that the reduction in net metabolic cost in the beginning of the powered condition is related to the reduced EMG activity in the m. soleus and the m. gastrocnemius as skeletal muscle contraction needs chemical energy which must be complemented by aerobic and anaerobic metabolic processes [18]. This reduced muscular activity seems to exceed the higher muscular activity in the small tibialis anterior muscle.

Following the initial reductions in metabolic cost, subjects further adapted during the 24 min powered condition. Only small kinematical changes occurred from the beginning of the adaptation to the adapted period but it is unlikely that these would strongly affect net metabolic cost. Furthermore, exoskeleton power was identical for the beginning of the adaptation and the adapted period. Despite these 'external similarities' between the beginning of the adaptation and the adapted period, the reduction in net metabolic cost increased from 9 % to 16 %. This 'internal difference' is assigned to reduced muscular activity: the mean RMS for the m. tibialis anterior, the m. soleus and the m. biceps femoris was significantly reduced. Subjects seemed to learn how to walk with the exoskeleton, neuromotor adaptations occurred and muscular activity was reduced when subjects adapted to the new walking gait. This adaptation phenomenon is too short to call it a learning effect but the mechanism shows similarity with children which evolve from gross activation toward a more economical muscle activity production when they learn to walk [19] or the reduction in muscle activity while learning a task [20,21].

Our results show that pneumatic ankle-foot exoskeletons can induce large reductions in metabolic cost in a short time period. Subjects seem kinematically adapted to walking with the exoskeleton within 4 min as only small differences in ankle and knee kinematics were found between the beginning of the adaptation and the adapted period. In accordance with our hypothesis subjects were metabolically adapted to our ankle-foot exoskeleton after 18.5 ± 5.0 min. Although it is possible that longer adaptations would cause further reductions in metabolic cost, all subjects seemed adapted within the

24 min powered condition, as no subject showed the lowest metabolic cost in the last 2 min. This adaptation time is consistently shorter than with exoskeletons using a proportional EMG feedback control mechanism [5–7,11], as they generally need adaptation times up to 90 min. The reduction in metabolic cost of 16 % during the adapted period compared to the unpowered condition is in agreement with the reduction of similar ankle-foot exoskeletons with ankle extension assistance during push off [8,9]. It is clear that a comparison with walking with normal shoes would reduce the reported reductions in metabolic cost. Still, the results of Malcolm et al. [9] indicate that walking with a powered exoskeleton can still be more economically than walking with normal shoes. Furthermore, we assume that even higher reductions in metabolic cost can be achieved if material properties of the exoskeleton can be optimized or by minimizing the negative power of the exoskeleton in the initial swing phase which counteracts the m. tibialis anterior.

Although some research suggests that the proportional EMG control method has greater potential for high reductions in metabolic cost [4], we found reductions that are greater than those reported for exoskeletons with an EMG based control method [5–7]. We believe that ankle-foot exoskeletons based on EMG have great value for neurological studies because of the direct interaction with biological EMG, in elderly because of the relatively normal gait kinematics [4], or for non-cyclical movements (e.g. stand-up and sit-down). Despite these benefits, we believe that this control algorithm includes a disadvantage: if a subject reduces EMG activity as a result of the exoskeleton, this leads to reduced assistance of the exoskeleton while our ankle-foot exoskeleton preserves a constant intermittent assistance across strides. Because of the low number of subjects and the lack of a genuine control condition, some caution is required in the interpretation of our results and more research is necessary to draw strong conclusions. Still, we want to highlight the possibilities of ankle-foot exoskeletons with kinematic control that causes ankle extension assistance during push-off: apart from the above mentioned fast adaptations and high reductions in metabolic cost, they have a relatively simple design, do not need complicated fine-tuning and can be applied quickly. These exoskeletons do not depend on the user's own EMG and could therefore be used in populations with altered muscular activity to restore normal gait [22]. The adaptation to exoskeletons in these populations needs to be studied in more detail as they probably need longer adaptations [23] or custom control [22].

In summary, when subjects walk with our ankle-foot exoskeleton with ankle extension assistance during push-off, an almost immediate reduction in metabolic cost of 9% is accompanied by a reduction in muscular activity of the plantar flexor muscles. This reduction increases to 16 % after 18.5 min when subjects are metabolically adapted. The difference between the beginning of the adaptation and the adapted period is the result of a neuromotor adaptation which reduces muscular activity in almost all leg muscles and makes subjects walk more economically with the exoskeleton. As in our hypothesis this highlights the interaction between the neuromotor adaptation and the metabolic adaptation. Because of the simple control method in combination with the fast adaptation and the high reductions in net metabolic cost, we want to highlight the potential of an ankle-foot exoskeleton with ankle extension assistance during push off. This understanding of the adaptation to walking with a powered exoskeleton with optimal kinematical actuation [9] is one of the many necessary steps in the evolution toward exoskeletons in daily life [2].

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Conflict of interest statement

None.

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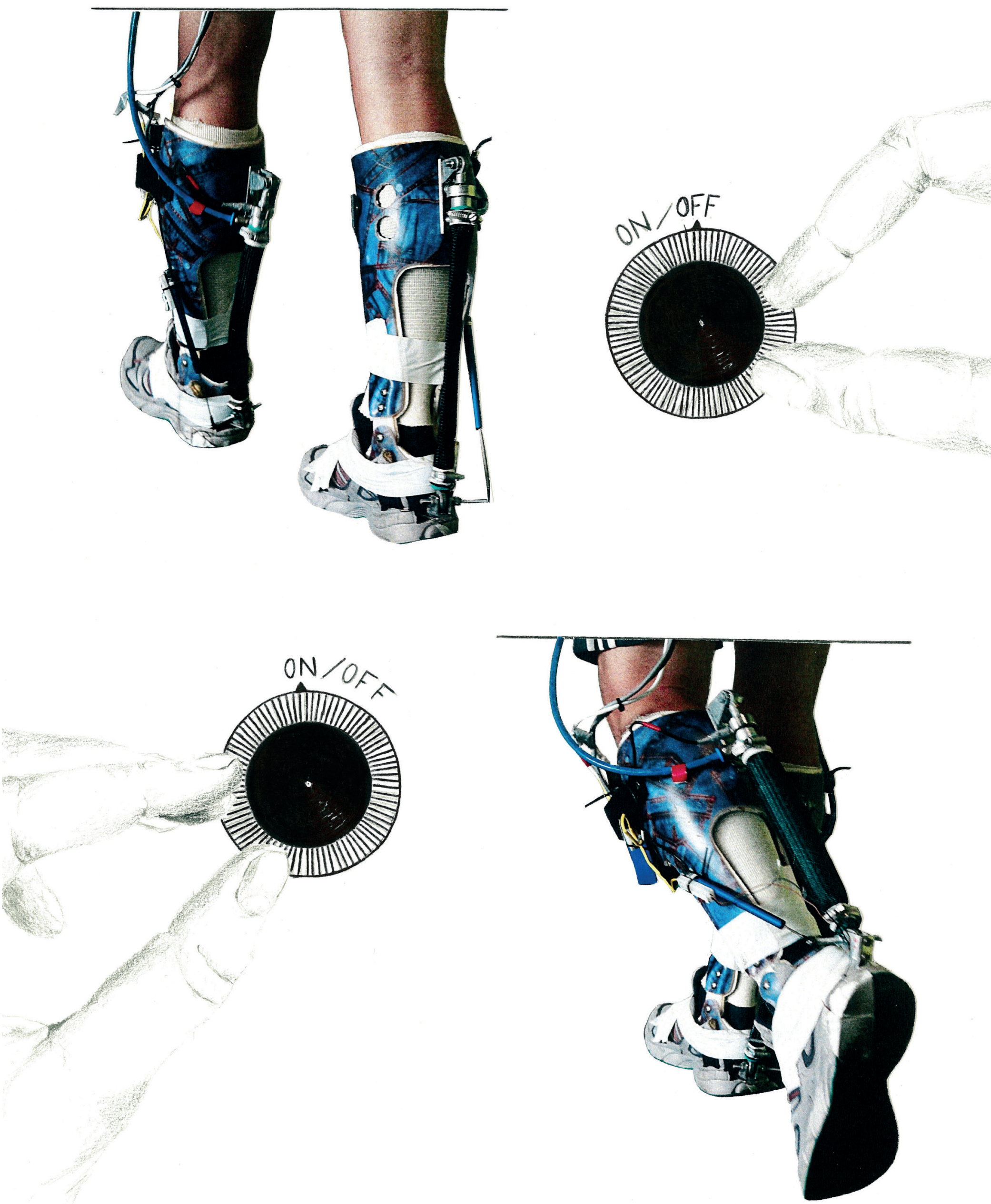
CHAPTER 3



MINIMIZING THE METABOLIC COST OF WALKING BY OPTIMIZING ACTUATION TIMING AND AVERAGE POWER OF ANKLE EXOSKELETONS

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*equal contribution

(first draft of manuscript to be submitted in 2015)

1. Abstract

Powered ankle-foot exoskeletons can reduce the metabolic cost of walking. It is unclear if the metabolic cost of exoskeleton walking can further be reduced with optimizing the actuation onset timing of the exoskeleton assistance and the average positive exoskeleton ankle mechanical power over a stride. We conducted a parameter sweep experiment with an ankle-foot exoskeleton, powered with pneumatic muscles that deliver plantar flexion power during the push-off. We used 4 actuation onset timings, close to the expected optimum (with onset at 36, 42, 48 and 54 % of the stride), and 3 exoskeleton ankle power levels (with total average positive exoskeleton ankle power over a stride, summed for both legs, of 0.2 $W{\cdot}kg^{-1}$, 0.4 $W{\cdot}kg^{-1}$ and 0.5 $W{\cdot}kg^{-1}$). Our data indicated that an actuation onset timing around 40% of the stride and average exoskeleton positive power around 0.4 $W{\cdot}kg^{-1}$ is optimal, leading to reductions in the metabolic cost of 21% for powered exoskeleton walking versus walking with the exoskeleton without assistance and a reduction of 12% versus normal walking without an exoskeleton. The assistive mechanism leading to these reductions seemed to include reducing the muscular activity of the plantar flexors during the push-off and probably assisting leg swing initiation, which reduced muscular activity in more proximal muscles. These results emphasize the importance of optimizing exoskeleton actuation timing and average exoskeleton power to reduce the metabolic cost of exoskeleton walking below the metabolic cost of normal walking.

2. Introduction

Walking is the most frequent way of locomotion in humans (Mitchell, 2006). Despite the high efficiency due to the interchange between potential and kinetic energy (Cavagna and Heglund, 1977), humans consume a considerable amount of metabolic energy during walking. This metabolic energy mainly results from the muscle work necessary for the push-off in the ankle joint (Umberger, 2010) to compensate for the energy loss during the step-to-step transition (Donelan et al., 2002a), for swinging the legs (Doke et al., 2005) and to support body weight (Grabowski, A. et al., 2005; Umberger, 2010). Other contributors to the metabolic cost of walking are the costs associated with active lateral stabilisation (Donelan et al., 2004) and swinging the arms (Collins et al., 2009; Meyns et al., 2013; Ortega et al., 2008).

Exoskeletons that act in parallel with the ankle joint and assist plantarflexion can reduce the amount of energy required for walking, which is called the metabolic cost of walking (Collins et al., 2015; Malcolm et al., 2013; Mooney et al., 2014a). At the moment, the assistive mechanism of these assistive exoskeletons is not fully understood. Several studies found alterations in the walking pattern and the muscle activity of other muscles than those around the ankle joint during plantar flexion assistance (Collins et al., 2015; Galle et al., 2013; Galle, Malcolm, Derave, et al., 2015; Koller et al., 2015; Malcolm et al., 2013), suggesting that the assistive mechanism of ankle-foot exoskeletons is more complex than the exoskeleton replacing some of the biological ankle joint work. The push-off in walking is related to reducing the collision cost during the step-to-step transition (Donelan et al., 2002a; Kuo, 2002) and Malcolm et al. (Malcolm et al., 2013) indeed suggested that plantar flexion assistance with an exoskeleton reduced the cost to redirect the centre of mass during the step-to-step transition and thus the collision cost (Kuo, 2002). Other research relates an impulsive ankle push-off to powering leg swing (Lipfert et al., 2014), which suggests that plantar flexion assistance with an exoskeleton might reduce leg swing cost.

Several studies focused on delivering positive ankle joint work with an exoskeleton during the push-off (Ferris, Czerniecki, et al., 2005; Sawicki and Ferris, 2008), leading to the first study showing a reduction in the metabolic cost compared to normal walking with a device that was tethered to off-board hardware (Malcolm et al., 2013) and the first study showing a reduction in the metabolic cost compared to normal walking with a fully autonomous device (Mooney et al., 2014a). A possible approach for optimizing exoskeleton assistance in order to further reduce the metabolic cost of exoskeleton walking is to optimize the exoskeleton assistance. Malcolm et al (Malcolm et al., 2013) showed that timing of exoskeleton actuation onset is an important determinant that influences the metabolic cost of walking. They found a U-shaped pattern when plotting metabolic cost versus actuation onset timing with an optimum around 40% of the stride. When other studies were compared (Norris, Granata, et al., 2007; Sawicki and Ferris, 2008), they seemed to match the U-shaped relationship. However, their method to change actuation timing between conditions also resulted in differences in average exoskeleton ankle power between conditions. The exoskeleton delivered much more ankle power in conditions with an early actuation onset compared to conditions with a late actuation onset, which could have influenced the results. Therefore, it would be valuable to study the effect of actuation onset timing when average exoskeleton ankle power is fixed, similar to what is done with a unilateral prosthesis emulator (Malcolm et al., 2015).

It seems evident that the average exoskeleton ankle power is also an important determinant of the metabolic cost of exoskeleton walking. However, different studies used different amounts of average exoskeleton ankle power to assist human walking (Malcolm et al., 2013; Mooney et al., 2014a; Sawicki and Ferris, 2008, 2009c), frequently between 50 and 80% of what the biological ankle delivers during unpowered exoskeleton walking (Sawicki and Ferris, 2008, 2009c). Based on the simplest walking model (Kuo, 2002) it can be expected that increasing exoskeleton ankle power will keep reducing the mechanical energy requirements for walking until subjects walk without metabolic energy cost. However, there have been experiments with exoskeletons (Collins et al., 2015) and simulation studies (Robertson et al., 2014; Zelik et al., 2014) with different parameters that are related to walking assistance, suggesting that “more is not always better”. Indeed, it seems evident that very low exoskeleton ankle power will only have a minimal effect, while from a certain amount of exoskeleton ankle power it will become difficult to walk with the exoskeleton. Research with a unilateral exoskeleton (Jackson and Collins, 2014) and with a unilateral prosthesis emulator (Caputo and Collins, 2014a) suggests an exponential relationship between average exoskeleton ankle power and the metabolic cost of walking, where the metabolic cost further reduces even when the device delivers almost twice the biological ankle joint work.

Focussing on the relationship between actuation onset timing, average exoskeleton ankle power and metabolic cost while measuring parameters associated with ankle push-off, collision cost and leg swing cost, insight into the assistive mechanism of exoskeletons and the effect on overall walking dynamics can be improved. The overall goal of this study is to optimize exoskeleton assistance, by optimizing exoskeleton actuation onset timing and average exoskeleton ankle power by means of a tethered and powered plantar flexion assisting exoskeleton, in order to further reduce the metabolic cost of walking with ankle-foot exoskeletons. Our aims are (1) to study the effect of actuation timing when average exoskeleton ankle power is fixed, (2) to study the effect of average exoskeleton ankle power at fixed actuation timings and (3) to improve the understanding of the human-exoskeleton interaction by analysing neuromechanics in order to find an explanation for the reductions in metabolic cost.

Concerning aim (1) we expect a second order relationship between the reduction in metabolic cost and the actuation onset timing with an optimal timing around 40% of the stride time based on our previous exoskeleton study (Malcolm et al., 2013). Concerning aim (2) we expect an exponential relationship for the reduction in metabolic cost versus average exoskeleton power assistance based on unilateral exoskeleton (Jackson and Collins, 2014) and prosthesis experiments (Caputo and Collins, 2014a). Concerning aim (3) we expect that exoskeleton assistance will influence the overall walking pattern, which will influence collision cost and/or leg swing cost and related parameters as both are suggested to be related to an impulsive ankle push-off (Kuo, 2002; Lipfert et al., 2014).

3. Methods

Subjects

Fourteen female subjects participated in the experiment but due to technical failure of the exoskeleton (2 subjects), drop-out (1 subject) and error in the synchronization signal (1 subjects), 10 subjects were retained (10 women; age 23 ± 1.2 y; weight 61.0 ± 4.5 kg; height 168.1 ± 5.2 cm; European shoe size 38.6 ± 0.8) and completed all experimental conditions with successful data collection. We selected female subjects of normal height and weight to be able to use one exoskeleton for the entire experimental population and to be able to apply large relative amounts of exoskeleton power because of their rather low body weight. None of the subjects had prior experience with exoskeleton walking. All participants signed a written informed consent and the experimental protocol was approved by the ethical committee of the Ghent University Hospital.

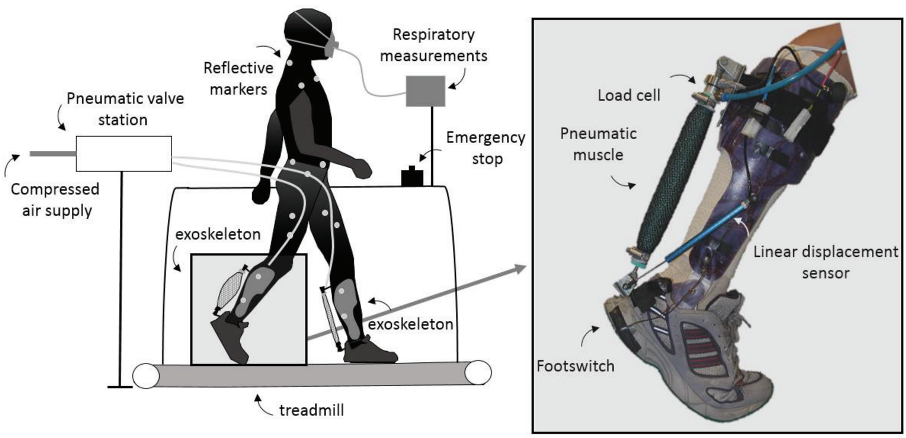


Fig. 1
A subject walking with the ankle-foot exoskeleton during the experiment with some of the measures that were done. The pneumatic muscles assist plantar flexion during the push off for the left and right leg. A load cell that is mounted proximal to the pneumatic muscle measures the tension force of the pneumatic muscle and a linear displacement sensor that is connected between the foot and the shank segment is used to measure ankle kinematics in real-time during walking.

Exoskeleton

The exoskeleton (Fig. 1) was similar to the exoskeleton that was used in our previous studies (Galle et al., 2014; Galle, Malcolm, Derave, et al., 2015; Malcolm et al., 2013). It consists of an ankle-foot orthosis with a hinge at the ankle joint and pneumatic muscles of 0.27m at the posterior side that are connected between the foot segment and the shank segment. These pneumatic muscles ‘contract’ when they are inflated with compressed air and cause ankle plantar flexion. The locations of the pneumatic muscles insertions were individually adjusted such that the pneumatic muscles allow exactly 15 degrees of dorsiflexion in passive state. The exoskeleton fitted in standard sport shoes where footswitches (Multimec 5E/5G, Mec, Ballerup, Denmark) were built in. A load cell (100 Hz; 210 Series, Richmond Industries Ltd., Reading, United Kingdom) was connected between the orthosis and the pneumatic muscle to measure pneumatic muscle force of the left and right leg. A linear displacement sensor (100 Hz; SLS130, Penny&Giles, Christchurch, United Kingdom) was connected between the foot and shank segment of the exoskeleton to measure ankle joint kinematics in real-time during walking. Total weight of one exoskeleton with all sensors that were carried by the user was 0.8 kg.

Actuation timing and average exoskeleton ankle power control

Based on footswitch signals of the previous stride, exoskeleton actuation of the next stride was controlled using fixed percentages of the stride time with a custom made feedforward algorithm in Labview (National Instruments, Austin, TX, USA), similar to our previous studies (Galle et al., 2013, 2014; Galle, Malcolm, Derave, et al., 2015; Malcolm et al., 2013). This software allows to impose a specific actuation onset timing and ending timing for the pneumatic muscles when subjects walk with the exoskeleton.

To control the average amount of positive exoskeleton ankle power over a stride, exoskeleton ankle power was measured in real-time during walking. Based on a prior calibration with motion capture, ankle joint angular velocity and moment arm of the pneumatic muscle force of the right leg were measured with the linear displacement sensor that was mounted between the heel and shank. Custom software in Labview allowed to calculate exoskeleton ankle power in real-time during walking. Moment arm and pneumatic muscle force were multiplied to estimate exoskeleton torque and exoskeleton torque was multiplied with ankle joint angular velocity to calculate exoskeleton power. Positive exoskeleton power was averaged over the duration of 10 strides and summed for the left and right leg to estimate average positive exoskeleton power over a stride. An iterative learning algorithm was developed to adjust air pressure for the pneumatic muscles (which influences pneumatic muscle force) during walking so as to deliver constant average exoskeleton positive power during walking.

Experimental conditions

A total of 14 conditions was done in each session. This consisted of a NormalWalking condition, in which subjects walked with normal shoes, without an exoskeleton, a ZeroWork condition, in which subjects walked with the exoskeleton but without assistance of the pneumatic muscles, and 12 powered conditions in which the exoskeleton assist plantar flexion. The twelve powered conditions were based on a two dimensional parameter sweep of four actuation timings and three average power levels. Four timing values where applied, with onset at 36, 42, 48 and 54 % of the stride and ending at 64 % of the stride. These values were chosen to be close to the optimal actuation timing of previous research (Malcolm et al., 2013). Three exoskeleton power conditions were applied, with total average positive exoskeleton ankle power over a stride (summed for both legs) of 0.2 W·kg^{-1}, 0.4 W·kg^{-1} and the maximal amount of average power that was possible with our exoskeleton and our control method. These values where chosen within the limits of our exoskeleton testbed and coincide with app. 50 and 100% of what the biological ankles delivers during unpowered exoskeleton walking (Sawicki and Ferris, 2008, 2009c). However, it was impossible to reach more then 0.2 W·kg^{-1} total average positive exoskeleton ankle power in the latest actuation timing due to bandwidth limitations of the pneumatic actuators and lower amounts of power in this timing appeared to be not relevant for practical use. It was therefore chosen to include the NormalWalking condition, the ZeroWork condition and only 10 powered conditions in our analysis (Fig. 5): three actuation timings (named respectively *Earliest, Early* and *Late*) for which three power levels were applied (named respectively *LowPower, MediumPower* and *HighPower*) and a fourth timing (*Latest*) where only low power was delivered by the exoskeleton (*LowPower*) due to the bandwith limitations of the pneumatic muscles.

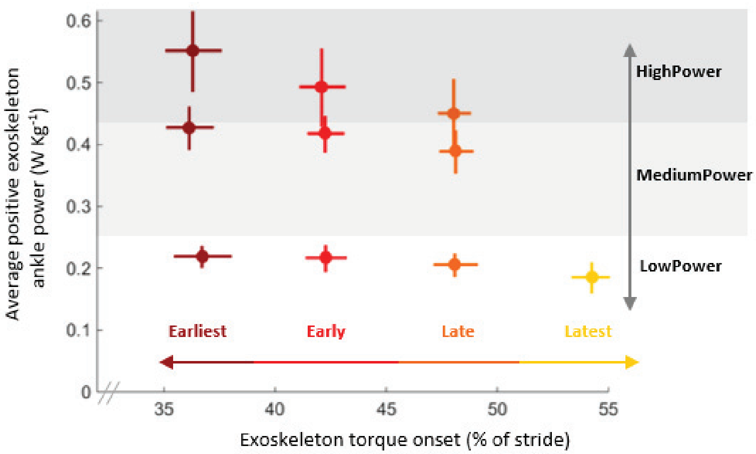


Fig. 2
Parameter sweep of exoskeleton actuation timing (which is the onset timing of the exoskeleton torque) and exoskeleton power (average positive exoskeleton ankle power). Dots in the figure are population averages for the four actuation timings (*Earliest – Early – Late – Latest*) and three exoskeleton power levels (*LowPower – MediumPower – HighPower*) and resulted in 10 powered conditions that are clearly distinct from each other. Actuation timing is expressed as a percentage of the stride time and exoskeleton power is expressed as the average positive exoskeleton power over a stride, summed for both legs. Horizontal and vertical lines around the dots indicate the standard deviations for our 10 experimental conditions for timing (horizontal) and power (vertical).

Experimental protocol

Subjects came in for a habituation session and an experimental data collection session with roughly one week in between. In the habituation session subjects learned to walk with the exoskeleton and where able to get used to the experimental set-up. In this habituation session subjects performed all 12 powered exoskeleton conditions, a zero-work condition (with the exoskeleton but without actuation of the pneumatic muscles) and a normal walking condition (without the exoskeleton and with their own sport shoes) during walking on a treadmill at 1.25 $m \cdot s^{-1}$. All conditions lasted 3 min and apart from the normal walking condition (which was always first or last due to the time to don and doff the exoskeleton) the exoskeleton conditions were done in randomized order with 2 min of rest in between. This habituation session was also used to fine-tune air pressure of the pneumatic muscles to reach the imposed average exoskeleton power for each subject individually. On the second day, in which experimental data were collected, subjects performed the same randomized walking protocol as in the habituation session but conditions lasted for 4 min and a standing rest condition, in which subjects stood still for 4 min, was done at the start of the experiment. The air pressure of the pneumatic muscles at the beginning of each powered condition was set to be the same as the air pressure that was used in the habituation session to reach the imposed average exoskeleton power with the iterative learning controller. Due to differences between subjects (both in walking dynamics and in body weight) this was done individually and resulted in homogeneous average exoskeleton power levels (*LowPower*, *MediumPower*, *HighPower*) and only small fluctuations in air pressure throughout each condition.

Data collection

In the habituation session only measurements of the displacement sensor, footswitches and load cells data were collected to allow measurements of exoskeleton power in real-time and adjust air pressure of the pneumatic muscles in order to reach the desired levels of exoskeleton power with the iterative learning controller. During the experimental data collection on a second day, exoskeleton sensors (footswitches, displacement sensors and load cells) were measured continuously for all conditions and used to control exoskeleton actuation timing and levels of exoskeleton power. Subjects wore a face mask which was connected with a gas analysis system that measured O_2 consumption and CO_2 production continuously during all conditions (Cosmed, K4b2, Rome Italy).

Full body 3D kinematics were recorded by 51 reflective markers (4 on each foot, 2 on each exoskeleton foot segment, 2 on each exoskeleton ankle joint, 6 on each exoskeleton shank segment, 2 on each knee joint, 4 on a plate connected to each thigh, 6 on the hip, and 5 on the torso) recorded with 14 infrared camera’s (200 Hz; Pro Reflex, Qualisys AB, Gothenburg, Sweden) collected with Qualisys software. Surface EMG of the m.soleus, m. gastrocnemius medialis, m. tibialis anterior, m. vastus lateralis, m. rectus femoris, m. biceps femors and m. gluteus maximums of both the left and right leg were measured with bipolar surface electrodes and wireless transmitters (1000 Hz; ZeroWire, Noraxon, Scottsdale, AZ, USA). Electrodes were placed in accordance with SENIAM guidelines (Hermens et al., 2000) with some restrictions for electrode placement of the m. soleus and m. gastrocnemius due to their location under the exoskeleton. Marker data and surface EMG data were collected simultaneously for 10s during the 3th and the 4^{th} minute of each condition, which includes around 9 full strides. In the few cases where subjects did not walk in a normal way in the 4^{th} minute (e.g. due to touching their face or scratching their body), data of the 3^{th} minute were used instead.

Data processing

Although exoskeleton power was calculated in real-time during the experiment using the linear displacement sensor, the exoskeleton power calculations in the results were based on calculations after the experiment with motion capture. Moment arm of the pneumatic muscles was calculated as the shortest perpendicular distance between the pneumatic muscle and the ankle joint axis. This moment arm was multiplied with pneumatic muscle force and divided by body weight to calculate exoskeleton torque during walking. This exoskeleton torque was multiplied with ankle joint angular velocity to calculate exoskeleton power.

Metabolic power of walking was calculated based on O_2 consumption and CO_2 production of the last 2 min of each condition (min 2 to 4), using a standard equation (Brockway, 1987). Net metabolic power was calculated by subtracting resting metabolic power from the standing rest trial from gross metabolic power during walking and dived by body weight.

Visual 3D software (C-Motion, MD, USA) was used to calculate kinematics. Marker positions and pneumatic muscle force data were filtered with a Butterworth low-pass filter with a cut-off frequency of 12 Hz . An 8 segment model (2 feet, 2 shanks, 2 thighs, 1 hip and 1 torso) was used to calculate body centre of mass and joint angles. For the kinematic model, weight of the arms and head were added to the torso and weight of the foot segment and the shank segment of the exoskeleton were added to the foot and shank in the exoskeleton conditions. Heel contact and toe-off were detected automatically based on kinematics (O'Connor et al., 2007).

Centre-of-mass work was calculated to improve understanding of overall (exoskeleton) walking mechanics, similar to other exoskeleton and prosthesis studies (Caputo and Collins, 2014a; Collins and Kuo, 2010; Collins et al., 2015; Jackson and Collins, 2015; Malcolm et al., 2015). Centre-of-mass velocity and acceleration were respectively calculated by taking the first and second derivative of centre-of-mass position in a treadmill belt fixed reference system. Ground reaction force was estimated based on the centre-of-mass acceleration and body mass. Total body centre-of-mass power was then calculated as the dot product of centre-of-mass velocity and the ground reaction force. As we did not use an instrumented treadmill we could only calculate centre-of-mass power using a combined limb method

(Donelan et al., 2002b). Centre-of-mass power was averaged for the positive and negative bursts around the double stance phase and during the single stance phase. During the double stance phase these were referred to as phases of positive and negative power and during the single stance phase they were referred to as the rebound and preload phase in accordance with the functional phases of the walking stride of Kuo et al. (Kuo et al., 2005).

Body-centre-of-mass calculations, kinematics and exoskeleton kinetics were time normalized from heel contact to the next heel contact and averaged for the left and right leg to calculate average plots.

Surface EMG data were band pass filtered (50-450 Hz) rectified and followed by a moving root mean square with a window of 100 ms. All EMG values were normalized to the peak value of the ZeroWork condition. Depending on the functional role of the muscle groups, peak values for functional phases of the gait cycle or stride averages were calculated. For the m. soleus and the m. gastrocnemius the maximal value during the push-off (which was set to be between 30 % and 70 % of the stride) was calculated. For the m. gluteus maximus, the m. vastus lateralis and the m. biceps femoris the maximum during the collision phase (which was set to be between 95 % and 15 % of the stride) was calculated and for the m. tibialis anterior and the m. rectus femoris the stride average was calculated.

Actuation timing refers to the onset of exoskeleton torque, was expressed as a percentage of the stride time and calculated based on the maximum of the second derivative of the unfiltered exoskeleton torque (i.e. when torque abruptly increases). Exoskeleton power refers to the average positive exoskeleton ankle mechanical power for a stride. This was calculated by averaging the positive exoskeleton ankle power over a stride for the left and right leg and was summed for both legs to calculate total average positive exoskeleton ankle power.

Statistics

To evaluate steady state in the metabolic measurements a repeated measures analysis of variance was done between the subsequent 30s averages of the 4 min conditions. To test the effect of actuation timing and exoskeleton power on the metabolic cost, a repeated measures analysis of variance was done, only with the exoskeleton conditions (10 powered conditions and a ZeroWork condition). If the P value was at or below 0.05, pairwise comparison was done by means of a paired t-tests with a Šídák-Holm correction for multiple testing. We decided that NormalWalking should not be part of the analysis that focusses on actuation timing and exoskeleton power as these variables are not manipulated during walking without exoskeleton. To compare exoskeleton conditions with NormalWalking, repeated measures analysis of variance was done with the NormalWalking condition included. Paired t-tests were used to check for differences between exoskeleton conditions and NormalWalking, with a Šídák-Holm correction for multiple testing. All statistics were done with Matlab (MathWorks, Natick, MA, USA).

Surface fit

To study the effect of changes in actuation timing and level of exoskeleton power on the metabolic cost of exoskeleton walking, we calculated a three-dimensional regression for actuation timing, exoskeleton power and the resulting metabolic cost (expressed as the reduction in metabolic cost versus the ZeroWork condition). Actuation timing and exoskeleton power were our independent variables and the metabolic cost was the dependent variable. This regression analysis was done based on the population averages for the 10 powered exoskeleton conditions and the ZeroWork condition, leading to 11 conditions for which we calculated the actuation timing, the exoskeleton power and the resulting metabolic cost. In the ZeroWork condition the average positive exoskeleton ankle power is 0 and the reduction in metabolic cost (when compared with the ZeroWork condition) is also 0. Therefore, the surface has to go through 0 for the metabolic reduction when average power is 0, independent of the actuation timing. The ZeroWork condition was therefore added for every actuation timing (*Earliest, Early, Late, Latest*) so that the surface fit would include the ZeroWork condition for all actuation timings. As a result, the regression analysis was done with 14 conditions. We did not perform a linear regression as we had specific hypotheses on the relationship between actuation timing, average positive exoskeleton ankle power and metabolic cost of walking. In the regression formula we included an exponential relationship for the relationship with average power and a second order relationship for actuation timing. The relationship for actuation timing was also multiplied with exoskeleton power to make sure that ZeroWork leads to a reduction in metabolic cost close to 0. Actuation timing onset (Ton) was expressed as a percentage of the stride time (e.g. 43%) and averaged total positive exoskeleton ankle power (Pavg) was expressed as the average positive exoskeleton power per stride, summed for both legs and normalized for body mass (e.g. 0.2 $W \cdot kg^{-1}$). Delta metabolic cost (ΔE) was the reduction in metabolic cost versus the ZeroWork condition (e.g. 0.51 $W \cdot kg^{-1}$). The following formula was used:

$$\Delta E = -a + b \cdot \exp(c \cdot Pavg) - d \cdot Ton \cdot Pavg + e \cdot Ton^2 \cdot Pavg$$

The same formula was also used to establish the relationship between surface EMG data (peak values and average values), actuation timing and exoskeleton power and for the centre of mass calculations. Regression analysis was done with Matlab (MathWorks, Natick, MA, USA) and *P* at or below 0.05.

4. Results

Exoskeleton mechanics

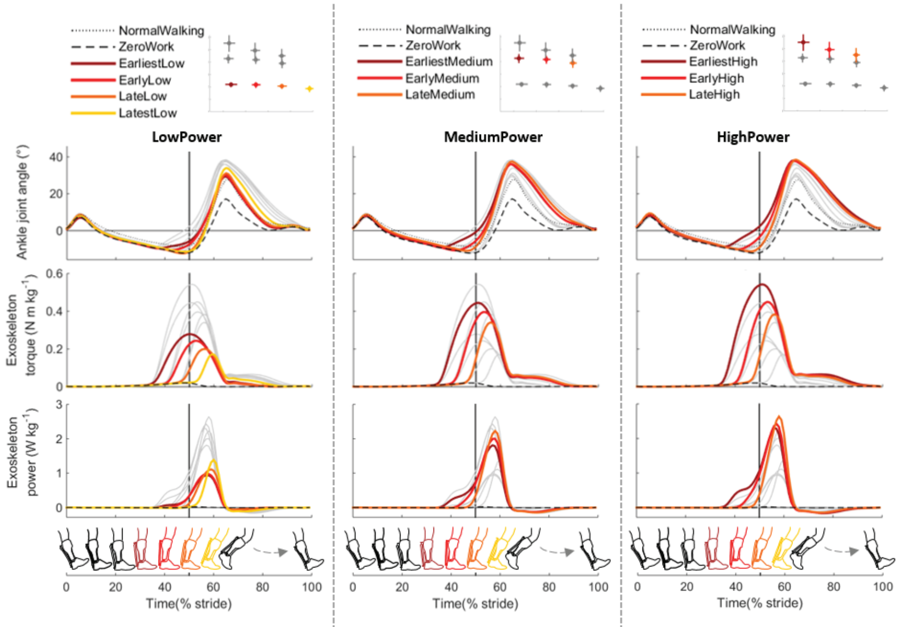


Fig. 3

Ankle joint angle, exoskeleton torque and exoskeleton power of the 10 powered conditions, the ZeroWork condition and the NormalWalking condition. Data are population averages, normalized from heel contact to heel contact. All graphs include all powered conditions in grey and the *LowPower* conditions in colour (in the left figures), the *MediumPower* conditions in colour (in the middle figures) and the *HighPower* conditions in colour (in the right figures). The smaller plots in the right corners show which conditions from the parameter grid in Fig 2 that are shown in colour. In each figure the differences between the different colours represent an effect for the actuation timing (when average exoskeleton power is fixed) and differences between figures for lines with the same colour represent effects of exoskeleton power (when actuation timing is fixed).

Subjects walked in 10 powered exoskeleton conditions, a ZeroWork condition and a NormalWalking condition on treadmill. Observation of the distribution of the exoskeleton torque onset timing, further often referred to as actuation timing, and average positive exoskeleton ankle power over a stride, which was summed for both legs and further often referred to as exoskeleton power, indicates that we successfully covered the desired parameter space (Fig. 2) and that the powered conditions were clearly distinguishable from each other. Average positive exoskeleton ankle power, summed for both legs, was 0.21 ± 0.02 W·kg^{-1} for the *LowPower* conditions, 0.41 ± 0.03 W·kg^{-1} for the *MediumPower* conditions and 0.50 ± 0.06 W·kg^{-1} for the *HighPower* conditions. Onset of the exoskeleton torque was at 36 ± 1% of the stride in the *Earliest* conditions, 42 ± 1% in the *Early* conditions, 48 ± 1% in the *Late* conditions and 54 ± 1% in the *Latest* condition. Actuation ending based on exoskeleton torque was at 64 ± 1% for all powered conditions.

Earlier actuation timings resulted in an earlier onset of pneumatic muscle force and hence an earlier onset of exoskeleton torque (Fig. 3). Increased air pressure of the pneumatic muscles increased exoskeleton peak torque in higher power conditions (Fig. 3). These alterations in exoskeleton torque influenced the ankle joint angle, with an earlier plantar flexion onset with earlier actuation timings and higher peak plantar flexion angles with more exoskeleton power.

Metabolic cost

No significant differences were found between the 30s averages for the metabolic cost in the last 2 min of each condition, suggesting that subjects reached a steady state in all conditions in the last 2 min. The metabolic cost of walking with the exoskeleton without assistance of the pneumatic muscles, the ZeroWork condition, was 4.03 ± 0.74 $W \cdot kg^{-1}$. Exoskeleton assistance reduced this metabolic cost for almost all powered conditions with reductions of to 21% for the *EarlyMedium* condition (Fig. 4). Metabolic cost for this condition was 3.16 ± 0.55 $W \cdot kg^{-1}$, actuation timing started at 42.3 ± 0.8 % of the stride and average positive exoskeleton power summed for both legs was 0.42 ± 0.03 $W \cdot kg^{-1}$. Due to the penalty of 11.6 ± 9.6 % when wearing the exoskeleton, significant differences for exoskeleton walking compared to normal walking where only found in this *EarlyMedium* condition, where metabolic cost was 12% lower compared to the NormalWalking condition (3.60 ± 0.74 $W \cdot kg^{-1}$).

The regression formula for the reduction in metabolic cost versus the ZeroWork condition (ΔE), based on the actuation onset timing (Ton) and the average positive exoskeleton power summed for both legs (Pavg) was expressed by a surface fit:

$$\Delta E = -10 + 10 \cdot \exp(0.89 \cdot Pavg) - 0.62 \cdot Ton \cdot Pavg + 0.0075 \cdot Ton^2 \cdot Pavg$$

This significant regression ($P \leq 0.05$) resulted in an adjusted R^2 of 0.84, indicating that this 2D non-linear model combining actuation timing and average exoskeleton power allows to estimate the reduction in metabolic cost when walking with our exoskeleton. Based on the surface plot a U-shaped pattern seems present for timing versus metabolic cost and also for exoskeleton positive ankle power versus metabolic cost (Fig. 6). This surface plot suggests a minimum in the metabolic cost when actuation timing is around 41% and exoskeleton power is around 0.4 $W \cdot kg^{-1}$, which is close to the *EarlyMedium* condition.

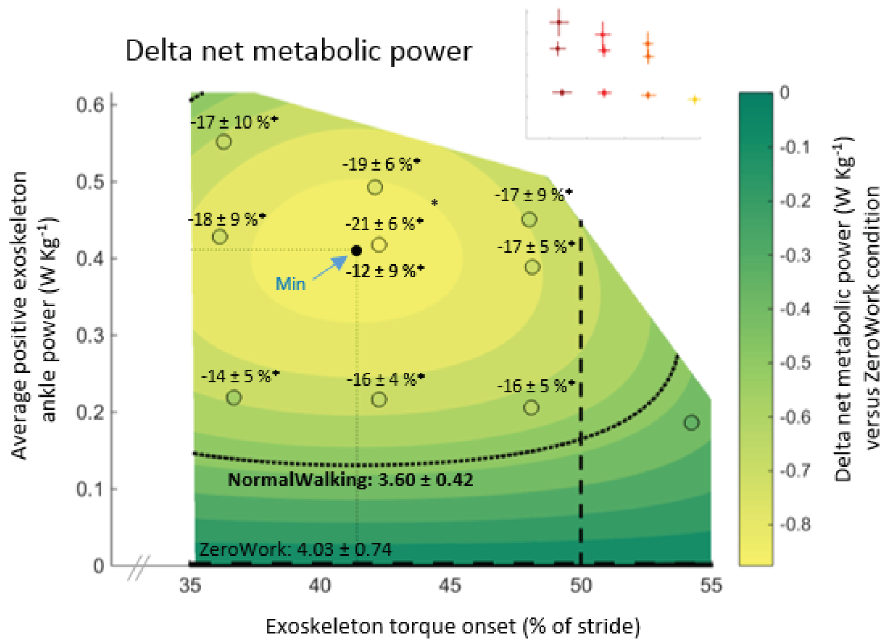


Fig. 4

Metabolic cost of exoskeleton walking for different actuation timings and average positive power levels. The metabolic cost is expressed as the reduction versus the ZeroWork condition for the 10 powered conditions. The small plot in the right upper corner shows the parameter grid of the 10 powered conditions that are represented in the figure, similar to fig. 3. The percentages above each condition are the reductions in metabolic cost versus the ZeroWork condition. The percentages underneath each condition (in bolt) are the reductions in metabolic cost versus NormalWalking. The green surface gradient is the result of a non-linear regression (see methods). The black dot with the blue arrow shows the predicted minimum based on the regression. The black dotted line shows the normal walking level (with a metabolic power of 3.60 ± 0.42 W·kg^{-1}) and the horizontal black dashed line shows the ZeroWork level (with a metabolic power of 4.03 ± 0.74 W·kg^{-1}). The vertical dashed black line indicates opposite heel contact. * indicate a sign. reduction versus ZeroWork (value above the condition) and ** indicate a sign. reduction versus NormalWalking (value underneath the condition and in bold) (*P* ≤ 0.05).

To establish the relationship between exoskeleton power and metabolic cost for fixed actuation timings, average positive exoskeleton ankle power was plotted against the reduction in metabolic cost versus the ZeroWork condition for each actuation timing separately (Fig. 5A). For each exoskeleton actuation timing, the *MediumPower* condition showed a higher reduction compared to the LowPower condition, indicating that more exoskeleton power leads to larger reductions in metabolic cost. However, the *HighPower* condition showed a smaller reduction in metabolic cost compared to the *MediumPower* condition for all actuation timings, suggesting that around these medium power levels at least a plateau was reached. For all actuation timings, the metabolic cost was lowest in the MediumPower conditions, with an average positive exoskeleton ankle power of 0.41 W·kg^{-1}.

To establish the relationship between actuation timing and metabolic cost for fixed levels of average exoskeleton power, actuation timing was plotted against the reduction in metabolic cost for each power level separately (Fig. 5B). A U-shaped pattern was found for the actuation timing for *LowPower*, *MediumPower* and *HighPower* conditions with the lowest metabolic cost for the *Early* timing with actuation onset at 41% of the stride time. The relationship between actuation timing and metabolic cost seems consistent across different amounts of exoskeleton power and also the relationship between exoskeleton power and metabolic cost seems consistent across different actuation timings.

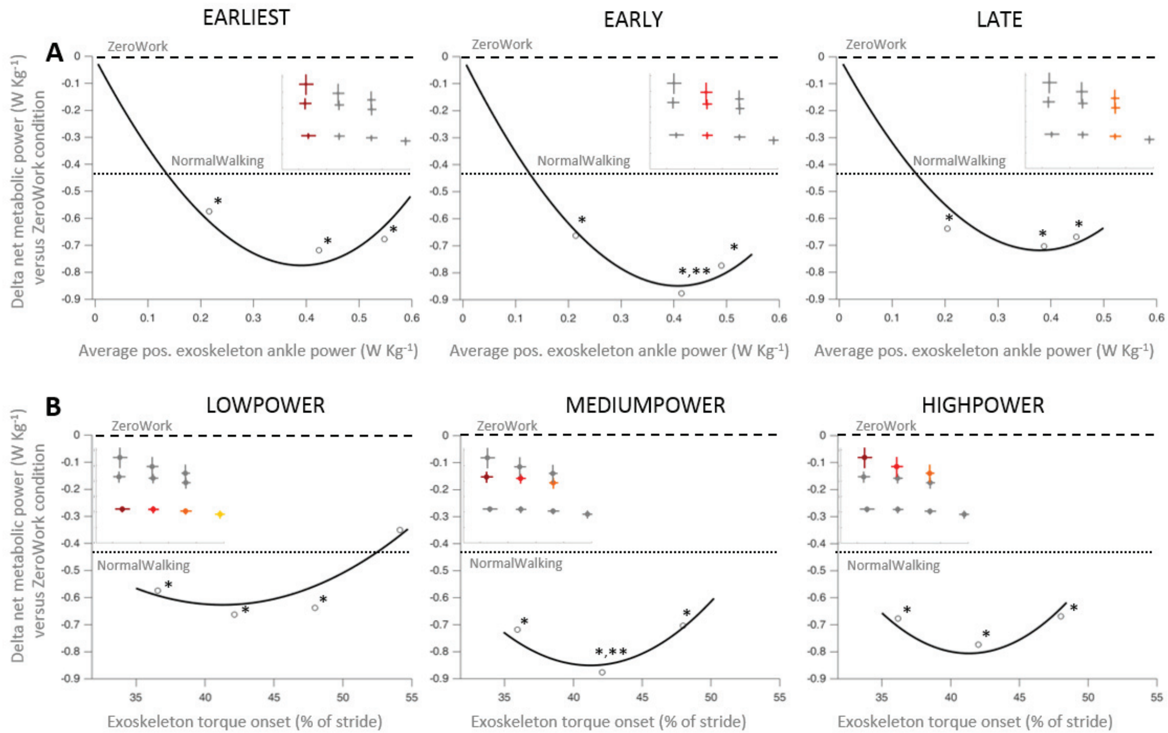


Fig. 5

Representation of the effect of average positive exoskeleton ankle power on metabolic cost (A) and the effect of actuation onset timing on metabolic cost (B). Data are identical to the data in figure 5, but separated for the effect of exoskeleton power (A) and actuation timing (B). Metabolic cost is expressed as the reduction versus the ZeroWork condition. The black solid line is the surface plot from Fig. 6, which is the result of a regression analysis, plotted for the range of the conditions in each graph. The black dashed line represents the metabolic cost for the ZeroWork condition and the black dotted line represents the metabolic cost for NormalWalking. All powered exoskeleton conditions have a metabolic cost below the ZeroWork condition and almost all powered conditions have a metabolic cost below the NormalWalking level. * indicate a sign. reduction versus ZeroWork and ** indicate a sign. reduction versus NormalWalking ($P \leq 0.05$).

EMG

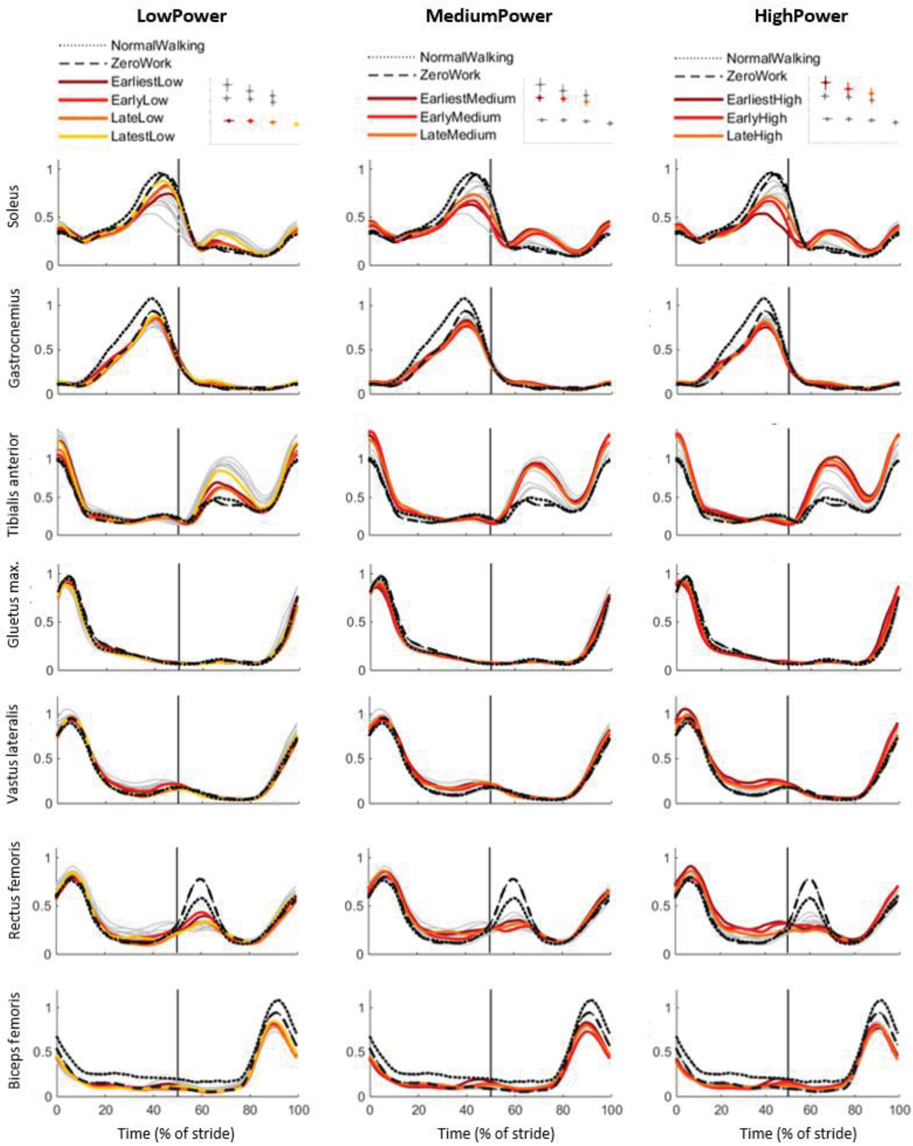


Fig. 6

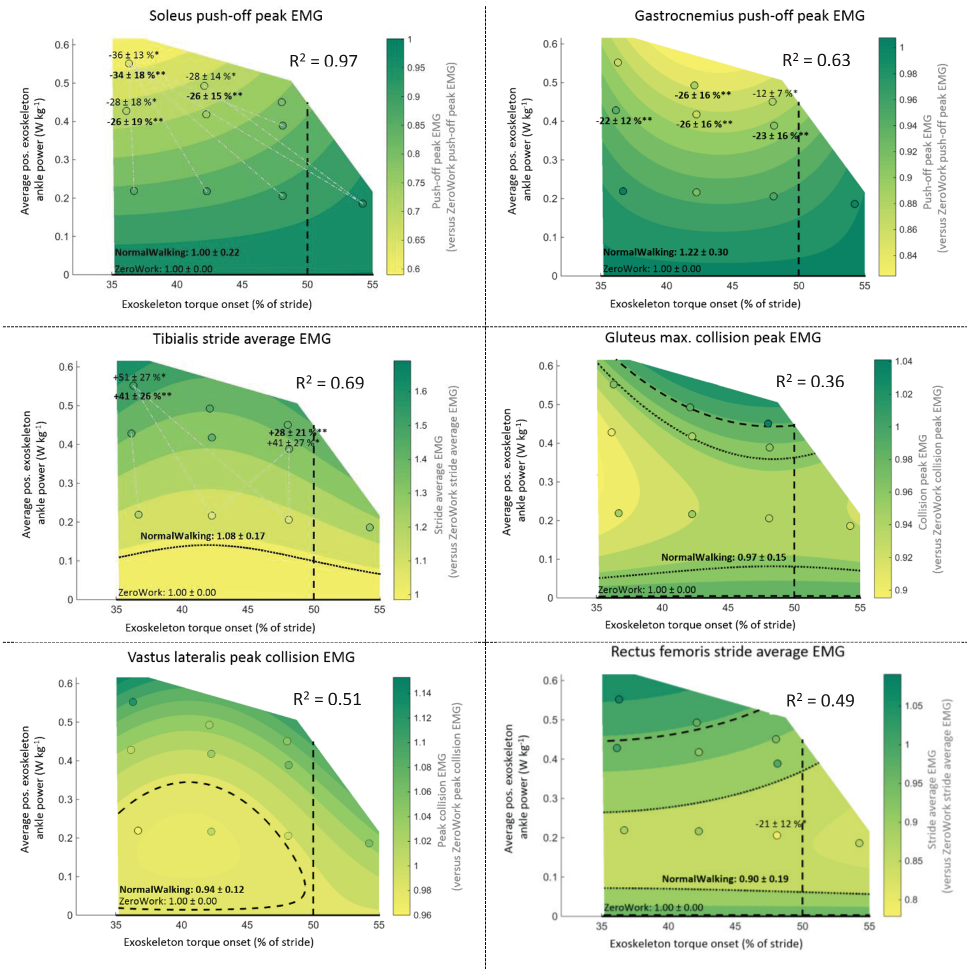
Electromyography for the most important leg muscles. Population averages for EMG curves of the lower leg muscles, averaged for the left and right leg and plotted from heel contact to heel contact. EMG's are normalized to the peak value for each subject in the ZeroWork condition. The black vertical line represents opposite heel contact. All graphs include all powered conditions in grey and the LowPower conditions in colour (in the left figures), the MediumPower conditions in colour (in the middle figures) and the HighPower conditions in colour (in the right figures). The smaller plots in the right corners show which conditions from the parameter grid in Fig. 4 that are shown in colour. In each figure the differences between the different colours represent an effect for the actuation timing (when average exoskeleton power is constant) and differences between figures for lines with the same colour represent effects of exoskeleton power (when actuation timing is constant).

Stride figures of EMG allows to understand the function of the specific muscles during exoskeleton walking and how EMG activity changes with exoskeleton assistance (Fig. 6). Exoskeleton assistance reduced m. soleus activity during the push-off. It should be noted that the increased muscle activity in the beginning of the stance phase and in the beginning of the swing phase could result from crosstalk with the tibialis anterior muscle. During the push-off however, tibialis anterior muscle activity is neglectible and therefore the soleus muscle activity during the push-off seems reliable. When the peak muscle activity was analysed, significant reductions in muscle activity of more than 30% versus the ZeroWork condition were found for earlier actuation timings and high levels of exoskeleton power. The significant regression analysis based on timing and power indicates that the largest reduction in m. soleus EMG could be found with an early actuation (Fig. 7). Soleus muscle activity showed also reduction versus the NormalWalking condition of more than 30% for earlier actuation timings and high amounts of exoskeleton power.

Also in the gastrocnemius muscle a reduction in the push-off peak was seen when compared with the ZeroWork condition (Fig. 6) . However, only for the *LateHigh* condition this resulted in a significant reduction of 12% versus the ZeroWork condition. Regression analysis indicated that the largest reduction in m. gastrocnemius activity could be found for an optimal timing (41%) with even higher amounts of exoskeleton power than the maximum that we applied (Fig. 7). Several powered conditions showed peak EMG activity that was more than 20% lower compared to the NormalWalking condition.

Tibialis anterior muscle activity shows a large increase in the beginning of the swing phase (Fig. 6). Average EMG activity over a stride was increased with more than 50% versus the ZeroWork condition for the *EarlyHigh* condition. The regression analysis indicates that high amounts of exoskeleton power increase the m. tibialis anterior activity (Fig. 7). The average EMG activity was increased with more than 30% for the *EarlyHigh* condition and more than 20% for the *LateHigh* condition when it was compared with NormalWalking.

For the m. gluteus maximus, EMG over a stride did not seem to be strongly influenced by exoskeleton assistance (Fig. 6). No significant differences were found when peak EMG activity during the collision phase was compared between conditions (Fig. 7). The regression was weak and based on small differences between conditions. Also for the m. vastus lateralis the effect of exoskeleton assistance on EMG was less clear (Fig. 6) and no significant differences were found between conditions when peak EMG activity during the collision phase was analysed (Fig. 7). Regression analysis showed only a weak relationship where muscle activity seems to increase with more exoskeleton power.



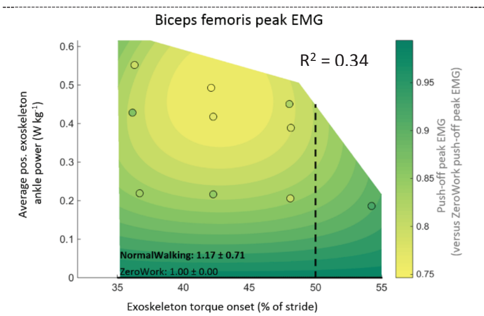


Fig. 7. EMG values based on the stride averages of Fig. 8 for the limb muscles in relation to actuation timing and exoskeleton power. Dots in the figures represent peak values during the push-off phase, peak values during the collision phase or stride averages, according to what is indicated in the figure and all compared to the ZeroWork condition. The percentages above each condition are the reductions versus the ZeroWork condition. The percentages underneath each condition (in bolt) are the reductions versus NormalWalking. The green surface is the result of a non-linear surface fit (see methods). R^2 values of the regression are represented for each figure. The black dotted line shows the normal walking level and the black dashed line shows the ZeroWork level. Gray dotted line shows sign. reductions between powered exoskeleton conditions. The vertical dashed black line indicates opposite heel contact. * indicate a sign. reduction versus ZeroWork (value above the condition) and ** indicate a sign. reduction versus NormalWalking (value underneath the condition and in bold) ($P \leq 0.05$).

For the m. rectus femoris, EMG profiles showed a large reduction in muscle activity in the swing phase (Fig. 6). However, averaged EMG over a stride showed only a significant reduction of 21% for the *LateLow* condition, when compared with the ZeroWork condition (Fig. 7). Regression analysis showed only a weak relationship with overall reductions that are larger for lower amounts of exoskeleton power. The m. biceps femoris showed a reduction in the powered conditions at the end of swing (Fig. 6). However, no significant differences were found when peak EMG in the collision phases was compared between conditions (Fig. 7). This might be due to the rather large standard deviations as average reductions where more than 20% for conditions with high amounts of power and actuation timings around the optimum. This resulted in a weak regression.

Centre-of-mass dynamics

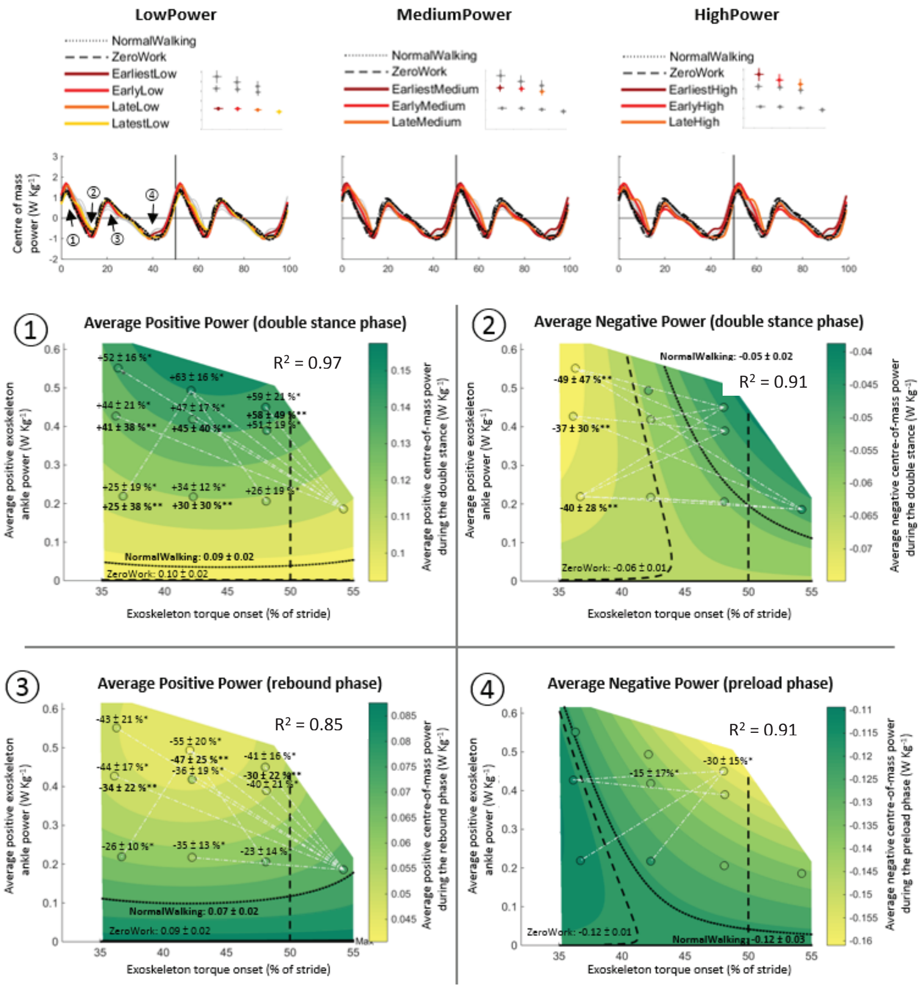


Fig. 8

Centre-of-mass power calculations. Curves on top of the figure are population averages for centre-of-mass power during the stride, represented from heel contact to heel contact. All graphs include all powered conditions in grey and the LowPower conditions in colour (in the left figure), the MediumPower conditions in colour (in the middle figure) and the HighPower conditions in colour (in the right figure). The smaller plots in the right corners show which conditions from the parameter grid in Fig. 4 that are shown in colour. The average centre-of-mass power for the positive burst during the double stance phase (①), the negative burst during the double stance phase (②), the positive burst during the single stance phase, also called the rebound phase (③) and the negative burst during the single stance phase, also called the preload phase (④), are shown for the first step in the left upper figure but can be repeated for the second step and for all conditions. These averaged bursts of positive and negative power are represented in the surface plots below, plotted against actuation timing and average power. The black vertical line represents opposite heel contact. Dots in the figures represent averages for positive and negative bursts of centre-of-mass power, according to what is indicated in the figure. The percentages above each condition are the reductions versus the ZeroWork condition. The percentages underneath each condition (in bolt) are the reductions versus NormalWalking. The green surface is the result of a non-linear surface fit (see methods). R^2 values of the regression are represented for each figure. The black dotted line shows the normal walking level and the black dashed line shows the ZeroWork level. Gray dotted line shows sign. reductions between powered exoskeleton conditions. * indicate a sign. reduction versus ZeroWork (value above the condition) and ** indicate a sign. reduction versus NormalWalking (value underneath the condition and in bold) ($P \leq 0.05$).

The average positive burst in centre-of-mass power during the double stance phase significantly increases with higher exoskeleton power levels (Fig. 8). The negative power during the double stance phase increases for early actuation timings but the overall effect is less clear. The average positive centre-of-mass power during the rebound phase significantly reduces for high levels of exoskeleton power. The negative centre-of-mass power during the preload phase increases for late actuation timings with high power levels but the overall effect is less clear.

5. Discussion

Main findings

The overall goal of this study was to optimize exoskeleton assistance, by optimizing exoskeleton actuation onset timing and average positive exoskeleton ankle power, in order to further reduce the metabolic cost of walking with ankle-foot exoskeletons. To study the effect of actuation timing and average exoskeleton power isolated from each other and to study the potential interaction between both we performed the most extensive parameter sweep reported in exoskeleton research until now. Most parameter sweeps use up to 7 powered conditions in one experiment (Caputo and Collins, 2014a; Jackson and Collins, 2014; Malcolm et al., 2015), while we performed 12 powered conditions from which 10 were retained.

Optimizing the actuation timing and the average exoskeleton ankle power resulted in a reduction in metabolic cost of 21 ± 6% compared to walking with the exoskeleton without assistance of the pneumatic muscles. This is the highest reported reduction for exoskeleton walking compared to walking with the exoskeleton without assistance as current reported reductions are 17% or lower (Malcolm et al., 2013; Mooney et al., 2014a; Sawicki and Ferris, 2008). However, due to the penalty of almost 12% when wearing the exoskeleton, the reductions in metabolic cost versus normal walking were smaller, with a reduction of 12% versus normal walking in the optimal exoskeleton condition. This is slightly higher than recently reported reductions for exoskeleton walking compared to normal walking, which are 10% or lower (Collins et al., 2015; Malcolm et al., 2013; Mooney et al., 2014a). Although recent findings indicate that reductions in metabolic cost of 7% versus normal walking can be found by assisting the isometric contraction during single stance, our results indicate that it is possible to obtain higher metabolic reductions by providing positive power assistance actuating predominantly during the double stance phase.

Our exoskeleton testbed was tethered to off-board hardware and power sources, while recent exoskeletons showed reductions in the metabolic cost for autonomous devices where all the hardware and power sources were carried by the user (Collins et al., 2015; Mooney et al., 2014a). Recent improvements in exoskeleton design reduced the penalty of wearing exoskeletons, even for fully autonomous devices (Asbeck et al., 2015; Collins et al., 2015; Mooney et al., 2014a), which suggests that reductions of more than 15% versus normal walking are possible when our exoskeleton assistance could be applied with an improved exoskeleton design. Despite the recent introduction of autonomous exoskeletons (Asbeck et al., 2015; Collins et al., 2015; Mooney et al., 2014a) our exoskeleton testbed is still useful as it allows to manipulate exoskeleton assistance parameters in a broad range, which is important to improve the understanding of the human-exoskeleton interaction.

Actuation onset timing

Our first aim focussed on the effect of actuation onset timing on the metabolic cost of walking. Based on our previous results (Malcolm et al., 2013), we expected a U-shaped relationship for the actuation timing and the reduction in metabolic cost with an optimum around 40% of the stride time. In this new experiment, average exoskeleton power was fixed between conditions and we still found a U-shaped relationship for metabolic cost with the lowest metabolic cost for the conditions with an actuation timing of 42% for all average exoskeleton ankle power levels. These results are consistent with the results of Mooney et al. (Mooney et al., 2014a) that used the same actuation timing for exoskeleton assistance, leading to high reductions in metabolic cost. The optimal timing corresponds with the predictions based on the simplest walking model (Kuo, 2002), where it was suggested that an impulsive ankle plantar flexion push just before contralateral heel contact is an energy efficient way to supply energy through mechanical work on the centre of mass that can reduce the collision loss at heel contact. The observed optimum is however much earlier than experimental results of a unilateral prosthesis emulator study, where an optimum was found at 52% of the stride or later (Malcolm et al., 2015). This difference could be due to the differences between exoskeleton walking and prosthesis walking, the unilateral assistance in the prosthesis study, the differences in the actuation profile or any combination of these. The observed optimum is also much later than the onset of torque in a passive exoskeleton with a mechanical clutch (Collins et al., 2015). However, the actuation with our exoskeleton is such that it delivers positive power mainly during the double stance phase, whereas the passive exoskeleton resulted in a reduction in metabolic cost mainly because of eccentric torque support during ankle dorsiflexion in the single stance phase. Both approaches result in significant reductions in metabolic cost, suggesting that a combined approach of passive torque support during the single stance phase, with powered assistance of the push-off in the double stance phase, might result in even higher reductions in metabolic cost.

Average positive exoskeleton ankle power

Our second aim focussed on the effect of average positive exoskeleton ankle power on the metabolic cost of walking. Based on the simplest walking model (Kuo, 2002), it is suggested that the metabolic cost should reduce with increasing exoskeleton power until walking requires no energetic cost. Other recent experiments with simulations and experiments (Collins et al., 2015; Robertson et al., 2014; Zelik et al., 2014) pointed several parameters of walking (assistance) where more is not always better, due to a trade-off between several factors that have a positive or a negative effect. Based on previous unilateral exoskeleton and prosthesis research (Caputo and Collins, 2014a; Jackson and Collins, 2014), we expected an exponential relationship for the reduction in metabolic cost and the average positive exoskeleton ankle power until at least 200% of the positive work that is delivered by the biological ankle joint during normal walking. Surprisingly, our data suggest a minimum in the metabolic cost around 0.2 $W{\cdot}kg^{-1}$ of average positive exoskeleton ankle power per leg, or a total of around 0.4 $W{\cdot}kg^{-1}$ for both legs, which is around 100% of the total biological positive ankle joint work during walking without exoskeleton assistance (Sawicki and Ferris, 2008, 2009c). Higher levels of average exoskeleton power resulted in lower reductions in metabolic cost, especially in the earlier actuation onset timing conditions. Due to the limitations of our hardware, it was not possible to apply higher amounts of power, especially in the later actuation onset timings. This might influence our results as Mooney et al. showed that high amounts of power can indeed result in high reductions in metabolic cost (Mooney et al., 2014a) although they did not measure exoskeleton mechanical power directly. It is hard to estimate how the relationship would evolve when higher amounts of power are applied in late actuation timings. The optimal average exoskeleton ankle power of 0.2 $W{\cdot}kg^{-1}$ per leg is much lower compared to the findings of a unilateral prosthesis emulator study or unilateral exoskeleton study (Caputo and Collins, 2014a; Jackson and Collins, 2014) where metabolic cost further reduced until at least 0.4 $W{\cdot}kg^{-1}$ per leg. However, differences in the actuation and the control and fundamental differences between prosthesis walking and exoskeleton walking might be the reason for the different results. Also, as they only applied unilateral assistance, the total amount of average exoskeleton ankle power was similar to our total applied exoskeleton power. It is possible that humans are capable of dealing with a specific amount of positive exoskeleton power, independent of if the power is applied on one or on both legs, but further research on differences between unilateral and bilateral assistance needs to clarify this.

Assistance parameters

There is still no real consensus on the 'optimal actuation' that results in the largest reductions in metabolic cost. The exoskeleton mechanical power combines the assistance of the device and the interaction with the user, and seems therefore the most important measure to express exoskeleton assistance. Our results indicate that the actuation onset timing and the average exoskeleton positive power are strong predictors of the metabolic cost of exoskeleton walking and the regression suggests that the largest reductions in metabolic cost could be found with an actuation timing of 41% and average exoskeleton power of 0.4 $W{\cdot}kg^{-1}$. However, we were unable to fully control the actual shape of the exoskeleton power curve, and it is possible that other actuation parameters determine the metabolic cost more strongly. While we focussed on timing and amount of push-off power, Collins et al. (Collins et al., 2015) showed how torque support earlier in the stance can also reduce the metabolic cost of walking. Jackson et al. (Jackson and Collins, 2014) developed an exoskeleton where torque and power can be varied in a more controlled way, which could be used to manipulate the exoskeleton power profile and further focus on the combined assistance of torque support during the stance and positive power assistance during the push-off.

Biomechanical explanations for the metabolic cost

Our results seem to suggest that exoskeleton average power is optimal around 0.4 $W \cdot kg^{-1}$, which is around 100% of what the biological ankle delivers during walking (Sawicki and Ferris, 2008, 2009c). This seems consistent with several studies that suggest that plantar flexor assisting exoskeletons reduce the metabolic cost of walking by replacing (some) of the biological muscle work in the plantar flexors (Gordon et al., 2006; Kao et al., 2010; Kinnaird and Ferris, 2009; Sawicki and Ferris, 2008, 2009c). This suggests that our exoskeleton replaces all positive biological ankle joint work during the push-off and seems to explain why an onset timing around 40% of the stride is metabolically optimal as this coincides with the onset of positive biological ankle power and why average positive exoskeleton power is optimal around 100% of the average biological positive power. Even the reductions in metabolic cost seem to match with the estimated cost of the plantar flexors during walking, which is 20 to 25% of the metabolic energy cost of walking (Sawicki and Ferris, 2008, 2009c; Umberger, 2010).

However, the assistive mechanism seems more complex than simply replacing all of the ankle joint positive work as the muscular activity of the plantar flexor muscles was reduced but still existing and plantar flexion assistance also increased muscular activity in the m. tibialis anterior. This increase in m. tibialis anterior activity results from the increased plantar flexion in the push-off, and should result in a metabolic penalty. It is than the question how plantar flexion assistance during the push-off still resulted in a reduction in metabolic cost of 20%?

A possible explanation is that the exoskeleton assistance, apart from reducing the plantar flexor muscle activity during the push-off, also influences other joints during walking and influences the overall walking pattern. We reported such a mechanism during uphill walking (Galle, Malcolm, Derave, et al., 2015) and previously found reductions in muscle activity in more proximal muscles during level walking (Galle et al., 2013). It seems likely that the exoskeleton then influences two important determinants of the metabolic cost of walking: the step-to-step transition cost (Donelan et al., 2002a) and the leg swing cost (Doke et al., 2005). The step-to-step transition costs depends on the negative work performed in the collision to redirect the centre-of-mass velocity from one ark to another, and the positive work to compensate for the energy loss (Donelan et al., 2002a). It seems like exoskeletons replace some of the push-off work, as seen in the reduction of the muscular activity of the soleus and gastrocnemius muscles. Based on the simplest walking model (Kuo, 2002), Malcolm et al. (Malcolm et al., 2013) suggested a mechanism in which exoskeleton assistance reduced the step-to-step transition cost by reducing the negative work performed in the collision phase. Although we were not able to calculate the collision cost using the individual limbs method (Donelan et al., 2002b), average negative center-of-mass power during the double stance phase seems not reduced in the powered conditions. Also on the muscular level we did not find reductions in muscular activity in the m. gluteus maximus, the m. biceps femoris, the m. vastus lateralis or the m. rectus femoris around the collision phase. This seems consistent with findings of prosthesis experiments, where prosthetic ankle push-off work increases push-off center-of-mass work but did not reduce collision work (Caputo and Collins, 2014a; Malcolm et al., 2015). This suggests that the assistive mechanism of the exoskeleton does seem to replace some of the push-off work but does not seem to be related to reducing the collision cost.

Recently, a study of Lipfert et al. (Lipfert et al., 2014) linked the impulsive ankle push-off in walking to initiation of leg swing. Indeed, we found increased plantar flexion and a higher plantar flexion velocity during the push-off and reductions in the m. rectus femoris during leg swing initiation. Somehow this also influences the m. biceps femoris activity at the end of swing when the lower leg and foot needs to be decelerated. Together, these point in the direction of a more passive leg swing during powered exoskeleton assistance. Similar results were found in a prosthesis study (Caputo and Collins, 2014a), where ankle push-off work was also related to assisting leg swing initiation.

When exoskeletons replace all the positive ankle joint work in the ankles during walking, maximal reductions in the metabolic cost of 20 to 25% are possible. However, our results indicate that plantar flexion assistance during the push-off reduces the plantar flexor muscle activity but also influences more proximal muscles by assisting leg swing initiation. Our results show how this combined approach can reduce the metabolic cost with more than 20% compared to unpowered exoskeleton walking. Recent findings from simulations and experimental work (Collins et al., 2015; Umberger, 2010) showed that the metabolic cost of the plantar flexors is not only related to the push-off but also to isometric work earlier in the stance. Collins et al (Collins et al., 2015) recently showed how reductions in metabolic cost of 7% are possible from torque support during the beginning of the stance phase. The combination of replacing ankle joint work during the push-off, assisting leg swing initiation and torque support during the beginning of the stance phase suggests that reductions in the metabolic cost of more than 25% are possible, especially as Collins et al (Collins et al., 2015) showed that their midstance torque support led to inefficient, rapid shortening of plantar flexor muscles during the push-off.

Conclusion

The overall goal of this study was to optimize exoskeleton assistance, by optimizing exoskeleton actuation onset timing and average exoskeleton ankle power, in order to further reduce the metabolic cost of walking with ankle-foot exoskeletons and improve insight into the assistive mechanism. We showed that reductions in the metabolic cost of 21% versus unpowered exoskeleton walking and 12% versus normal walking are possible by optimizing actuation timing and average exoskeleton power. Actuation timing showed an optimum around 40% of the stride and average exoskeleton power was optimal around 0.4 $W \cdot kg^{-1}$. The assistive mechanism leading to these reductions includes reducing the muscular activity of the plantar flexors during the push-off and it was suggested that it also included assisting leg swing initiation, which reduced muscular activity in more proximal muscles.

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CHAPTER 4

UPHILL WALKING WITH A SIMPLE EXOSKELETON: PLANTAR FLEXION ASSISTANCE LEADS TO PROXIMAL ADAPTATIONS

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Uphill walking with a simple exoskeleton: Plantarflexion assistance leads to proximal adaptations

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1. Abstract

While level walking with a pneumatic ankle-foot exoskeleton is studied extensively, less is known on uphill walking. The goals of this study were to get a better understanding of the biomechanical adaptations and the influence of actuation timing on metabolic cost during uphill walking with a plantar flexion assisting exoskeleton.

Seven female subjects walked on a treadmill with 15% inclination at 1.36m·s^{-1} in five conditions (4 min): 1 condition with an unpowered exoskeleton and four with a powered exoskeleton with onset of pneumatic muscle actuation at 19, 26, 34 and 41% of stride.

During uphill walking the metabolic cost was more than 10% lower for all powered conditions compared to the unpowered condition. When actuation onset was in between 26 and 34% of the stride, metabolic cost was suggested to be minimal. While it was expected that exoskeleton assistance would reduce muscular activity of the plantar flexors during push-off, subjects used the additional power to raise the body centre of mass in the beginning of each step to a higher point compared to unpowered walking. This reduced the muscular activity in the m. vastus lateralis and the m. biceps femoris as less effort was necessary to reach the highest body centre of mass position in the single support phase.

In conclusion, subjects can use plantar flexion assistance during the push-off to reduce muscular activity in more proximal joints in order to minimize energy cost during uphill locomotion. Kinetic data seem necessary to fully understand this mechanism, which highlights the complexity of human-exoskeleton interaction.

2. Introduction

Walking is the most common gait mode in humans. During level walking, up to 70% of total energy is interchanged between gravitational-potential energy and kinetic energy [1], resulting in a metabolically economic gait pattern at normal walking speeds [2]. Although walking is the most economic gait mode for steep uphill inclines (>15%) when compared with running and cycling [3], the body needs to be raised against gravity: external mechanical work increases [4] and mechanical energy exchange is less effective [5]. Thereby, the required amount of positive mechanical work produced in the ankle, knee and hip joint increases [6,7], which results in an increase in the muscular activity of the lower limbs. Consequently, the metabolic cost of walking increases drastically with gradient [8]. Therefore, reducing the metabolic cost of uphill walking seems relevant for rescue workers, soldiers and in recreational hiking because of the frequently long walks in hilly areas.

Powered ankle-foot exoskeletons with plantar flexion assistance can reduce the metabolic cost of level walking compared to unpowered walking [9-13]. Malcolm et al. [14] found that timing of plantar flexion assistance is a crucial element towards metabolic optimization: they were the first to find a 6% reduction in metabolic cost of walking with a powered exoskeleton compared to walking with normal shoes. While this exoskeleton was tethered to a power source, autonomous exoskeleton devices have also shown reductions in metabolic cost versus normal walking [15,16].

Uphill walking can also benefit from plantar flexion assistance: although the major increases in muscular activity, joint moments and joint powers are situated around the hip joint, the contribution of the ankle also increases [6,7,17]. Sawicki and Ferris [11] found a 10% reduction in metabolic cost for powered walking on a 15% inclination compared to unpowered walking. They used an exoskeleton with proportional m. soleus EMG control, where biological ankle work was partially replaced by exoskeleton mechanical work. However, Malcolm et al. [14] used an exoskeleton with footswitch control to show that an actuation timing solely based on biological muscle activation is not metabolically optimal for level walking. The mechanical power of this footswitch controlled exoskeleton increases rather than replaces biological ankle joint work, which is used to redirect the body centre of mass during the step-to-step transition [14]. Subjects seem to use a different approach to deal with additional plantar flexion power, depending on the control system of the exoskeleton. Therefore, it is unclear how plantar flexion assistance with footswitch control will affect the trajectory of the body centre of mass during uphill walking and how actuation timing will influence the metabolic cost. Because ankle power generation coincides with the metabolically optimal actuation timing during level walking [14] and ankle power generation initiates around 30% of the stride during uphill walking on a 15% inclination [6,7], we predicted that the lowest metabolic cost will be found with an actuation starting around 30% of stride during uphill walking on a 15% inclination. This falls between the actuation onset of proportional m.

soleus EMG control during uphill exoskeleton walking [11] and the metabolically optimal actuation timing during level walking with a footswitch controlled exoskeleton [14].

The first goal of our study is to get more insight into the biomechanical adaptations during uphill walking with a powered ankle-foot exoskeleton, given the promising results of this device during level walking [14]. Because there is no consensus on the metabolically optimal actuation timing during uphill walking, the second goal of our study is to explore the influence of actuation timing on metabolic cost.

3. Methods

Exoskeleton

Subjects wore a bilateral pneumatic ankle-foot exoskeleton that assists plantar flexion during push-off. More details can be found in previous publications [9,14]. Actuation onset timing was set at 19, 26, 34 and 41% of stride in the uphill walking conditions (respectively TIM 1, TIM 2, TIM 3 and TIM 4). These actuation timings were chosen to be between 18% [11] and 43% [14] of stride. Actuation ending was set to toe-off. A previous experiment with the same exoskeleton [19] showed that net exoskeleton mechanical power per stride is app. 0.25 $W \cdot kg^{-1}$ when actuation starts at 36% of stride during uphill walking without additional weights.

Subjects and Protocol

Seven female subjects (age 21 ± 0 years; body mass 60.5 ± 5.3 kg; height 167.9 ± 3.9 cm; means ± s.d.) signed an informed consent, approved by the ethical committee of the Ghent University hospital. After a standing rest condition (4 min), subjects performed a habituation (24 min) during level walking with a powered exoskeleton [9] at a treadmill speed of 1.36 ± 0.02 $m \cdot s^{-1}$. An uphill walking habituation would have induced fatigue and as uphill walking seems to be controlled by a modification of the level walking motor programme [6], we assumed that this would not influence the differences between the following uphill walking conditions. Subjects performed five randomized uphill conditions (4 min) at the same speed on a 15% inclination with 3 min of rest after every condition: one unpowered condition in which subjects walked with the exoskeleton without actuation of the pneumatic muscles and four powered conditions.

Data collection

O_2 consumption and CO_2 production were measured during the entire experiment (Oxycon Pro, Jaeger GMBH, Ho Germany) and 44 markers (five located on the feet, four on the shanks, four on the thighs, four on the pelvis, two on each pneumatic muscle, two on each ankle, two on each knee and one on the left and right greater trochanter) were used to record kinematics with motion capture with 11 cameras (200 Hz; Pro Reflex, Qualisys AB, Gothenburg, Sweden). EMG (200 Hz; Zerowire, Aurion, Milan, Italy) was measured with surface electrodes, placed in accordance with SENIAM guidelines [18] for the left and right m. tibialis anterior, m. soleus and medial m. gastrocnemius and the right m. vastus lateralis and m. biceps femoris. High speed cameras (200 Hz; Bassler, Ahrensburg, Germany) allowed to detect heel contact and toe-off timing. A load cell (200 Hz; A.L. Design, Buffalo, NY, USA) was connected to the right pneumatic muscle to measure force but post-experiment analysis indicated that the load cell got damaged during the experiment. Therefore, load cell data were only used to detect pneumatic muscle actuation duration. Previous experiments have shown that exoskeleton mechanical work during uphill walking (12%) represent around 30% of normal ankle joint work during uphill walking [19]. Motion capture, EMG, high-speed video and load cell measurements were recorded during 10s in the last min of each uphill walking condition.

Data processing

Metabolic cost was calculated as metabolic power based on 30s averages of O_2 and CO_2 measurements with the formula of Brockway [20] and normalized by body weight. Flow rates were averaged for min 2-4 in the uphill conditions and the metabolic power of the quiet standing trial (min 2-3) was subtracted from the metabolic power during uphill walking to calculate net metabolic power. The metabolic power for the eight consecutive 30s averages within each 4 min interval were compared to assure that steady state was reached in the last 2 min.

Stance and swing duration were obtained from high-speed video recordings with MaxTraq software (Innovation Systems, Columbiaville, MI, USA). Marker- and load cell data were filtered with a 2nd order Butterworth low-pass filter (cut-off 6 Hz) and lower limb joint angles were calculated in Visual 3D (C-Motion, MD, USA) using a seven segment model (two feet, two shanks, two thighs and a hip). The angle in normal upright posture was deducted from ankle, knee and hip angles so that 0° represents upright posture. A stick figure was made based on segment position every 20% of stride. Centre of mass (COM) of the pelvis segment was used as a proxy for body COM [21]. Body COM position was normalized to the position of the foot at right heel contact and multiplied with treadmill speed and stride time to produce an overground body COM trajectory from right heel contact to right heel contact. Because of the safety bars of the treadmill, hip markers were insufficiently tracked in one subject. EMG data were high-pass filtered with a 2nd order Butterworth filter (cut-off 10 Hz) and rectified. Because of occasional bad electrode adhesion, EMG measurements for the m. vastus lateralis, m. biceps femoris and m. tibialis anterior were based on six subjects. Moving root mean square (RMS) was taken, EMG RMS amplitude was normalized to the peak value of the unpowered uphill walking condition and data of both legs were ensemble averaged. EMG and kinematics were time-normalized from heel contact to heel contact.

Statistics

All statistical comparisons were done with SPSS Statistics 20 (IBM, Armonk, NY,USA). Repeated measures ANOVA ($P \leq 0.05$) with post-hoc comparison ($P \leq 0.05$) and Bonferroni correction were done to compare net metabolic power between consecutive 30s averages within the 4 min intervals and between all uphill walking conditions and to compare actuation onset and offset for the four powered uphill walking conditions. Paired samples t-test ($P \leq 0.05$) was done to compare the unpowered condition with powered condition TIM 3 for stance time and swing time. Every 10% of stride paired samples t-tests ($P \leq 0.05$) were done to compare the unpowered condition with the powered condition TIM 3 for EMG and kinematics.

4. Results

Within the 4 min uphill walking conditions, the subsequent 30s averages showed significant differences in net metabolic power in the beginning of the intervals ($P \leq 0.05$) but not between the last four 30 s averages, indicating steady state in the last 2 min of each interval. During walking with the exoskeleton (Fig 1A), plantar flexion assistance resulted in significantly lower net metabolic power (F = 55.776; $P \leq 0.05$) for all powered conditions (TIM 1: 9.86 ± 1.07 W·kg^{-1}; TIM 2: 9.79 ± 0.90 W·kg^{-1}; TIM 3: 9.79 ± 0.96 W·kg^{-1}; TIM 4: 10.03 ± 0.96 W·kg^{-1}) compared to the unpowered condition (11.13 ± 0.87 W·kg^{-1})(Fig 1B). There seems a U-shaped pattern in net metabolic power with a local minimum when actuation onset is between 26 and 34% of the stride.

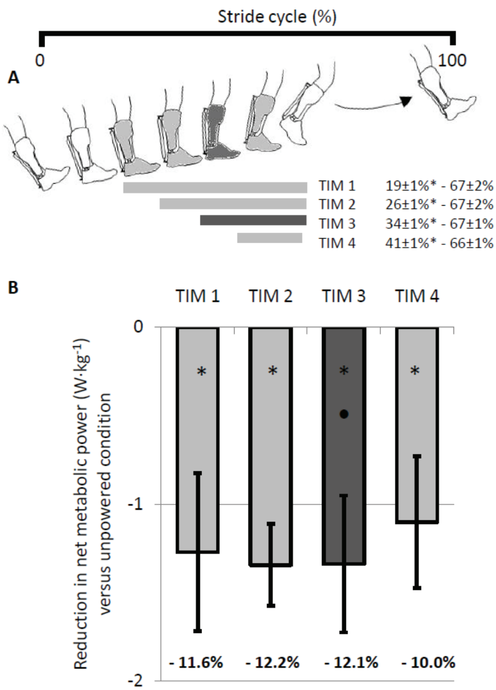

Fig. 1. Pneumatic muscle actuation duration (A) and metabolic cost (B) of the four powered uphill walking conditions.
A. Duration of the actuation of the pneumatic muscles for the four powered conditions (TIM 1, TIM 2, TIM 3, TIM 4). Bars represent actuation duration and percentages next to the bars are the actual actuation onsets and offsets in percentages of the stride time (mean ± s.d.). * indicates significantly different from all other conditions (ANOVA, $P \leq 0.05$).
B. Differences in net metabolic power of the four powered uphill walking conditions (TIM 1, TIM 2, TIM 3, TIM 4) versus the unpowered uphill walking condition with standard deviation. A negative value represents a reduction compared to the unpowered condition. Percentages under the bars represent reductions in net metabolic power versus the unpowered walking condition. * indicates a significant difference versus the unpowered walking condition (ANOVA, $P \leq 0.05$). • indicates a significant difference compared to the powered condition TIM 4 (ANOVA, $P \leq 0.05$).

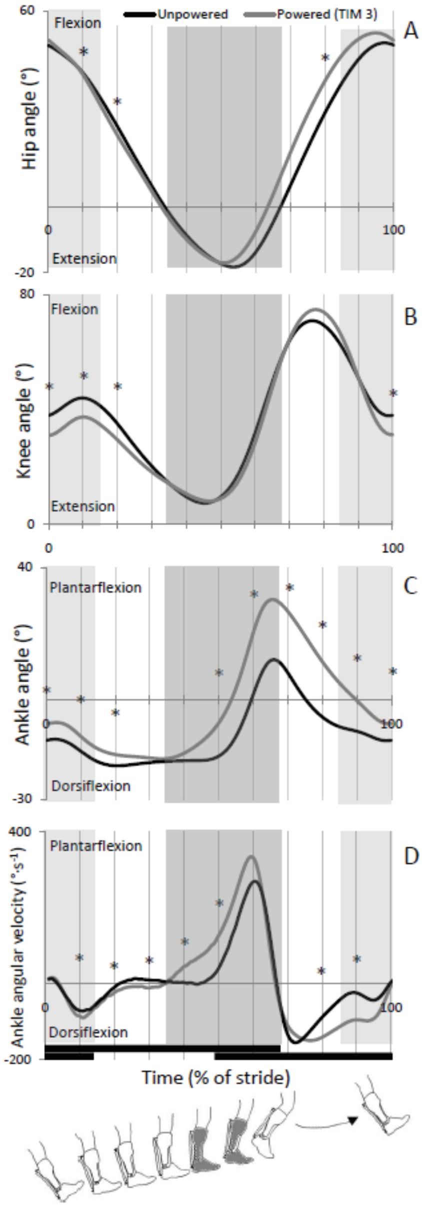

Fig. 2. Hip angle (A), knee angle (B), ankle angle (C) and ankle angular velocity (D) for the unpowered uphill walking condition (black line) and the powered uphill walking condition TIM 3 (grey line). Averaged values for seven subjects (A) or six subjects (B, C, D) are time normalized for the right leg from right heel contact to right heel contact. Dark grey shaded areas represent pneumatic muscle actuation period from the right leg (actuation from 34 to 67% of stride). Also the actuation period of the opposite leg is presented with a light grey shaded area at the beginning and the end of the stride. Black horizontal lines at the bottom show the stance phase duration of the left and right leg. ***** indicates significant differences (t-test, $P \leq 0.05$) every 10 % of stride between the unpowered condition (black) and the powered condition TIM 3 (grey).

Apart from reduction in metabolic cost, mechanical power input of the exoskeleton is important towards autonomous exoskeletons and previous experiments during level walking showed that a longer exoskeleton actuation duration produces more mechanical work [14]. During uphill walking, actuation onset showed significant differences ($F = 632.622$; $P \leq 0.05$) between all powered conditions ($P \leq 0.05$ for all) with the longest actuation duration in TIM 1 and the shortest in TIM 4. As the metabolic cost in TIM 3 was significantly lower compared to TIM 4, TIM 3 was chosen for further analysis as from a perspective of mechanical efficiency, the largest reduction in metabolic cost (compared to TIM 4) is accompanied with the least exoskeleton assistance (compared to TIM 1 and TIM 2).

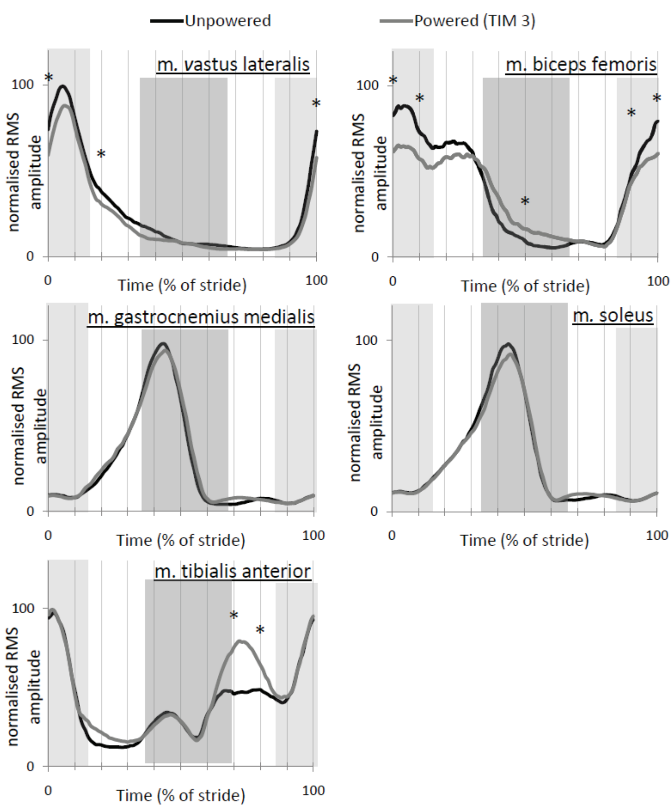

Fig. 3. Normalized RMS EMG amplitude of five muscles during a stride for the unpowered uphill walking condition (black line) and the powered uphill walking condition TIM 3 (grey line). Plots are means of the EMG RMS amplitude (with n being the number of subjects) for the m. vastus lateralis (n=6), m. biceps femoris (n=6), m. gastrocnemius medialis (n=7), m. soleus (n=7) and m. tibialis anterior (n=6), time normalized from heel contact to heel contact. RMS amplitudes are normalized to the peak of the unpowered curve. Dark grey shaded areas represent pneumatic muscle actuation period from the right leg (actuation from 34 to 67% of stride). Also the actuation period of the opposite leg is presented with a light grey shaded area at the beginning and the end of the stride. ***** indicate significant differences (t-test, $P \leq 0.05$) between the unpowered condition (black) and the powered condition TIM 3 (grey) every 10% of stride.

Both stance time (TIM 3: 0.69 ± 0.02 s; UNP: 0.69 ± 0.03 s) and swing time (TIM 3: 0.37 ± 0.03 s; UNP: 0.37 ± 0.02 s) showed no significant differences between TIM 3 and the unpowered condition. Kinematic differences were mainly found in the ankle, with altered angular velocity and more plantar flexion over almost the entire stride in TIM 3 (Fig. 2). The increased plantar flexion during the push-off is the result of the pneumatic muscle actuation and this effect remains during the first part of the swing phase. Differences between TIM 3 and the unpowered condition in the knee and hip joint are mainly situated in the beginning of the stride, with a more extended knee joint in the beginning of the stance phase in TIM 3, which is when the pneumatic muscle of the opposite leg is actuated.

Muscular activity in the m. soleus and the m. gastrocnemius shows no reduction in TIM 3 but muscular activity in the m. tibialis anterior is increased during the beginning of the swing phase (Fig. 3). In the m. vastus lateralis and the m. biceps femoris, significant reductions in muscular activity were found in the beginning and the end of the stride, when the pneumatic muscle of the opposite leg is actuated.

The body COM vertical position is significantly higher at the end of the pneumatic muscle actuation of each leg in TIM 3 (Fig. 4) compared to the unpowered condition. The stick figure in Fig. 4 allows to get a better interpretation of how pneumatic muscle actuation, alterations in body COM position and reduced muscular activity in the thigh muscles can be related to each other. The plantar flexion assistance results in increased plantar flexion which seems used to raise the body COM, which reduces the muscular activity in the m. vastus lateralis (knee extensor) and the m. biceps femoris (hip extensor) to raise the body COM to the highest point in the single support phase.

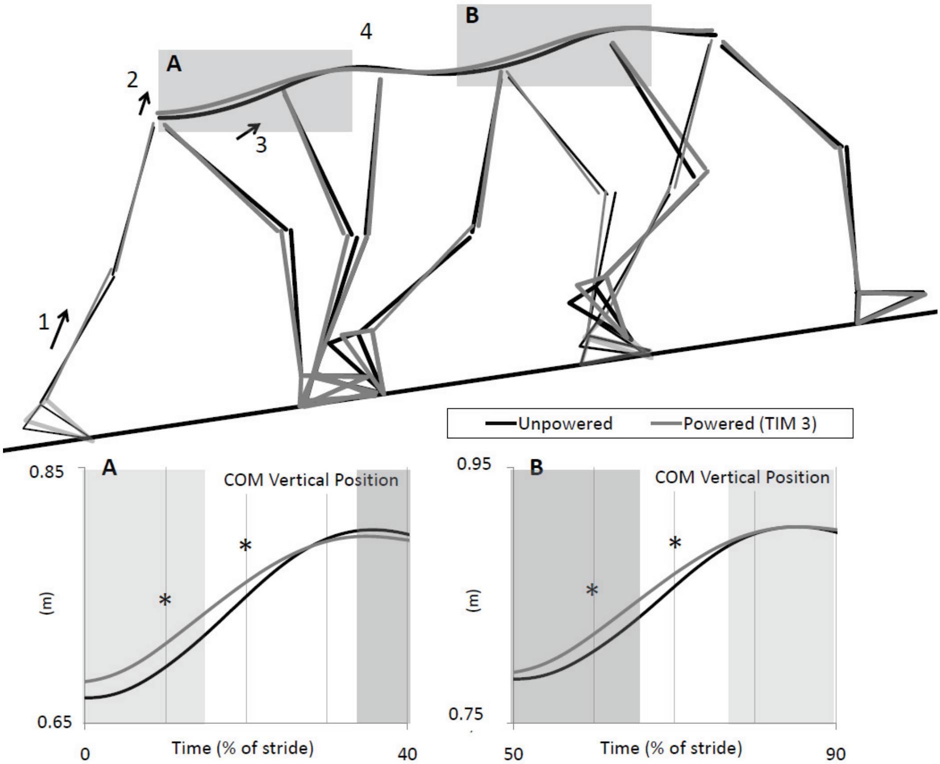

Fig. 4. Stick figure of the uphill walking movement and body COM vertical position (A and B) for the unpowered uphill walking condition (black line) and the powered uphill walking condition TIM 3 (grey line). The right leg is presented from right heel contact to right heel contact. Also the stance phase of the opposite leg is presented with thinner lines. In the beginning of the stride, the pneumatic muscle of the opposite leg is still actuated (1) and the additional plantar flexion is used to raise the body COM (2). Because of this higher position of the body COM, less effort is necessary (3) to raise the body COM to its highest point in the single stance phase (4). The same happens later in the stride when the right leg is actuated.
A and B: the vertical body COM position during uphill walking for the unpowered condition (black line) and the powered condition TIM 3 (grey line). Dark grey shaded areas represent pneumatic muscle actuation period from the right leg (actuation from 34 to 67% of stride). Also the actuation period of the opposite leg is presented with a light grey shaded area at the beginning and the end of the stride. Vertical position of the body COM was normalized from right heel contact to right heel contact. * indicate significant differences (t-test, $P \leq 0.05$) between the unpowered condition (black) and the powered condition TIM 3 (gray) every 10% of stride.

5. Discussion

During walking with a powered exoskeleton, the largest reductions in net metabolic power can be found with an onset in actuation timing in between 26% (TIM 2) and 34% (TIM 3) of the stride, resulting in reductions in net metabolic power of more than 12% compared to unpowered exoskeleton walking. While the relative reduction in metabolic cost for powered walking versus unpowered walking on an inclination is lower compared to the 17% reduction that was found during level walking, the absolute reduction of 1.34±0.39 $W{\cdot}kg^{-1}$ for TIM 3 is more than twice the reported reduction in metabolic cost during level walking with the same device [14]. Sawicki and Ferris [11] found a reduction in metabolic cost of 10% for powered compared to unpowered uphill walking. They used an exoskeleton with proportional m. soleus EMG control, resulting in an estimated actuation onset around 18% of stride. This is consistent with the reduction of 11.6±4.4% that was found in our experiment for an onset at 19% of stride during uphill walking.

The inclusion of a standard shoe condition would have made the results more meaningful but this comparison of powered versus unpowered uphill locomotion still added new insights to exoskeletal research. With the development of soft-suit exoskeletons [15], new designs [16] and energy-recycling approaches [22-24] the differences between unpowered walking and walking with standard shoes are becoming smaller and smaller and this results in autonomous exoskeletons that reduce metabolic cost below the level of normal walking [15,16].

Onset of biological ankle power generation can be estimated based on angular velocity because after initial plantar flexion, the remainder of the stance phase is characterised by an ankle plantar flexion moment in uphill and level walking [6,7,16]. During level walking, Malcolm et al. [14] found that the metabolically optimal actuation onset (43%) coincided with the onset of biological ankle power generation. However, the differences in onset of ankle plantar flexion velocity between the unpowered condition and the powered condition (Fig. 2) during uphill walking make it unlikely that the onset of ankle power generation is the critical determinant for metabolic optimization. Also, this indicates that the exoskeleton does not solely assists the plantar flexors and that the assistance that causes the reductions in metabolic cost involves a proximal mechanism.

A second indication towards a proximal mechanism is the fact that the additional plantar flexion was not used to reduce the biological plantar flexor activity during the push-off. It was expected that our exoskeleton would behave in a similar way as an EMG-steered exoskeleton [11] or a spring-like ankle-foot orthose [25] where the device is used to replace biological ankle muscle-tendon work rather than augment total ankle work. Our exoskeleton increased the plantar flexion during the push-off and the additional plantar flexion power seemed used to raise the body centre of mass. This resulted in a higher centre of mass position at the end of the pneumatic muscle actuation, which is the crucial phase where the body centre of mass has to be raised over the support foot (Fig. 5). This also changed the landing configuration of the opposite leg in the beginning of the stance phase towards a more extended position and reduced the muscular activity of the knee- and hip extensors (m. vastus lateralis and m. biceps femoris) when the body centre of mass needs to be raised through further knee- and hip extension towards its highest point in the single stance phase.

Previous studies showed that during uphill walking, the majority of work and muscular activity is situated around the hip joint [6,16]. We hypothesize that the locomotor system uses the additional plantar flexion during the push-off of the contralateral leg to reduce the effort in the joints of the support leg that are otherwise heavily loaded. This could result from the human nature to minimize energy cost during locomotion [26], thereby confirming the capability of the human system to adapt to forces that act in parallel to the joints. It seems like subjects used the additional plantar flexion to increase total ankle work in order to reduce work around the hip, which reduced the metabolic cost. Also during level walking with the same exoskeleton, a similar mechanism seemed present [14] as total ankle joint work was increased during powered walking compared to unpowered walking. This indicates that different exoskeleton control mechanisms can result in different approaches to deal with additional plantar flexion power, leading to reductions in metabolic cost. This highlights the complexity of the human-exoskeleton interaction and the need to get a better understanding of this interaction.

However, caution is required when interpreting our results. The relationship between EMG activity and force production is dependent on muscle length [27] and exoskeleton assistance can have negative consequences for muscle-tendon units [28], which in example can result in similar EMG activity but lower force production in the plantar flexor muscles in the powered condition. There seems to be a logical connection between ankle plantar flexion assistance and resulting changes in body centre of mass position and EMG, resulting in a reduction in metabolic cost but kinetic data, while taking into account individual leg function during double support [29], seem necessary to fully address our hypothesis and is a necessity for future work to better understand assistive locomotion.

In conclusion, uphill walking with a powered ankle-foot exoskeleton can reduce the metabolic cost compared to unpowered walking. Our results suggest that instead of replacing part of the plantar flexor muscles during the push-off a proximal mechanism causes the reduction in metabolic cost: the exoskeleton plantar flexion power is added to the biological ankle plantar flexion and used to raise the body centre of mass to a higher position, thereby reducing the muscular activity in the thigh muscles. While kinetic data seem necessary to support our hypothesis our results confirm the effectiveness of exoskeleton assistance during uphill walking and highlights the complexity of the human-exoskeleton interaction.

Conflict of interest statement

None declared.

Acknowledgements

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CHAPTER 5

ENHANCING PERFORMANCE DURING INCLINED LOADED WALKING WITH A POWERED ANKLE-FOOT EXOSKELETON

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ORIGINAL ARTICLE

Enhancing performance during inclined loaded walking with a powered ankle–foot exoskeleton

Samuel Galle · Philippe Malcolm · Wim Derave · Dirk De Clercq

1. Abstract

PURPOSE. A simple ankle-foot exoskeleton that assists plantar flexion during push-off can reduce the metabolic power during walking. This suggests that walking performance during a maximal incremental exercise could be improved with an exoskeleton if the exoskeleton is still efficient during maximal exercise intensities. Therefore, we quantified the walking performance during a maximal incremental exercise test with a powered and unpowered exoskeleton: uphill walking with progressively higher weights.

METHODS. Nine female subjects performed two incremental exercise tests with an exoskeleton: one day with (powered condition) and another day without (unpowered condition) plantar flexion assistance. Subjects walked on an inclined treadmill (15 %) at 5 km·h^{-1} and 5 % of body weight was added every 3 min until exhaustion.

RESULTS. At volitional termination no significant differences were found between the powered and unpowered condition for blood lactate concentration (respectively 7.93 ± 2.49 mmol·L^{-1}; 8.14 ± 2.24 mmol·L^{-1}), heart rate (respectively 190.00 ± 6.50 bpm; 191.78 ± 6.50 bpm), Borg score (respectively 18.57 ± 0.79; 18.93 ± 0.73) and $\dot{V}O_2$ peak (respectively 40.55 ± 2.78 ml·min^{-1}·kg^{-1}; 40.55 ± 3.05 ml·min^{-1}·kg^{-1}). Thus, subjects were able to reach the same (near) maximal effort in both conditions. However, subjects continued the exercise test longer in the powered condition and carried 7.07 ± 3.34 kg more weight because of the assistance of the exoskeleton.

CONCLUSIONS. Our results show that plantar flexion assistance during push-off can increase walking performance during a maximal exercise test as subjects were able to carry more weight. This emphasizes the importance of acting on the ankle joint in assistive devices and the potential of simple ankle-foot exoskeletons for reducing metabolic power and increasing weight carrying capability, even during maximal intensities.

2. Introduction

Exoskeletons that assist the lower limbs during locomotion have improved much in the last decade and experts in the field believe that they will soon have an important role in daily life (Ferris 2007). While most research is focused on technical enhancements, a quantitative evaluation of the effectiveness is often missing (Dollar and Herr 2008). The metabolic energy expenditure, often calculated as metabolic power ($W \cdot kg^{-1}$) based on oxygen consumption and carbon dioxide using a standard equation (Brockway 1987) and body weight normalization, is a key value in the evaluation of several exoskeleton devices (Galle et al. 2013a; Malcolm et al. 2013; Mooney et al. 2014; Norris et al. 2007; Sawicki and Ferris 2008, 2009a, 2009b; Wehner et al. 2013). Regardless of the functional goal of the device, reducing the metabolic power will improve the usability of the exoskeleton (Ferris et al. 2007). Accordingly, metabolic power can be considered a prime outcome when evaluating exoskeleton effectiveness, that can even be used to drive kinematic behavior with exoskeletons (Collins and Jackson 2013).

During walking, half of the positive joint work is done by the ankle during push-off (Winter 1983). Therefore, much potential is attributed to powered exoskeletons that assist ankle plantar flexion. Walking with powered exoskeletons with pneumatic muscles that assist plantar flexion during push-off results in reductions in metabolic power of 10 to 17 % compared to walking with an unpowered exoskeleton (without plantar flexion assistance)(Galle et al. 2013a; Malcolm et al. 2013; Norris et al. 2007; Sawicki and Ferris 2008, 2009a, 2009b). Despite the increased weight of the device, Malcolm et al. (2013) were the first to report a 6 % reduction in metabolic power for powered exoskeleton walking compared to walking with normal shoes if the actuation timing of the exoskeleton was optimal. While they showed that it was possible to reduce metabolic power below the level of normal walking, their device was not autonomous, meaning that it needed power and air supply and extensive hardware which was not carried by the user. However, Mooney et al. (2014) showed that it is possible to make a fully autonomous exoskeleton that assists plantar flexion and that can reduce the metabolic power of loaded walking with 8 % versus normal walking if the design of the device is altered in order to reduce distal mass. Furthermore, there is increasing progress in energy recycling approaches (Collins and Kuo 2010; Donelan et al. 2008; Li et al. 2009; Malcolm et al. 2013, Unal et al. 2012) and soft exosuits (Wehner 2013), which makes it likely that autonomous ankle-foot exoskeletons can become a permanent fixture in daily life.

Exoskeletons can be used as assistive devices for patients, *e.g.* to restore normal gait (Blaya and Herr 2004; Sawicki et al. 2006) but the applications for healthy subjects are less obvious. Because ankle-foot exoskeletons can reduce the metabolic power during walking, it should be possible to walk with higher external workloads (due to increasing slope, speed or carrying weights) when assisted by an exoskeleton, while the metabolic power requirements remain constant. At higher workloads it can be expected that ankle-foot exoskeletons have the potential to increase walking performance during a maximal incremental exercise test if two conditions are met: (A) the powered exoskeleton must still be effective in terms of reducing metabolic power during maximal exercise intensities and (B) it must be feasible for the user to walk with the exoskeleton with minimal encumbrance during maximal exercise intensities. If both these conditions are fulfilled, one could expect an external workload (slope, speed or carried weight) during walking with a powered exoskeleton that transcends the maximal achievable workload during walking without an exoskeleton.

However, we are not aware of any succesfull attempts in increasing walking performance during maximal exercise intensities as current research is mainly focused on submaximal intensities (Galle et al. 2013a; Malcolm et al. 2013; Mooney et al. 2014; Norris et al. 2007; Sawicki and Ferris 2008, 2009a, 2009b). Studying higher intensities is also useful for applications in specific populations (*e.g.* prolonged or loaded walking for soldiers and rescue workers) and will give more insight into human-exoskeleton interaction, *e.g.* on the efficiency and the consistency of the assistance over increasing intensities. Higher intensities would also contribute to taking exoskeletons out of their 'normal' environment and allow to use them as a tool to resolve fundamental questions in biomechanics, motor control and physiology, as suggested by Ferris et al. (2007). In example, the influence of muscle fatigue on overall fatigue could be studied by assisting or resisting specific muscles (Malcolm 2009) with an exoskeleton during exercise until exhaustion.

The pneumatic artificial muscles that are mostly used in ankle-foot exoskeletons have numerous benefits for assisted walking like the low weight to force ratio and their compliant behaviour (Daerden and Lefeber 2000). Although they need compressed air supply, which makes them less useful for daily life applications, they are frequently used in a lab environment to study general principles on human-exoskeleton interaction. Because they cannot achieve the high inflation and deflation frequency needed for running, higher intensities without drastically changing the walking pattern (Franz and Kram 2012; Harman et al. 2000; Lay et al. 2006; Lay et al. 2007) can only be achieved during uphill walking and by adding external weights (Kramer 2010). It seems reasonable that subjects can benefit from push-off assistance during loaded uphill walking as previous research showed that subjects can benefit from an exoskeleton during uphill walking (Sawicki and Ferris 2009a) and during load carrying (Mooney et al. 2014). Therefore, exoskeleton locomotion during maximal exercise intensities could be tested with the weighted walking test (Klimek and Klimek 2007) or a similar alternative. The weighted walking test is a method to assess aerobic power during walking: subjects walk on a treadmill with an inclination of 12 % at a speed of 1.8 $m \cdot s^{-1}$ and every 3 min 5 % of body weight is added until exhaustion. Klimek and Klimek (2007) showed that this is a valid alternative for a maximal cycling or running exercise test.

The aim of our study is to quantify the walking performance during a maximal incremental exercise with a simple powered ankle-foot exoskeleton with plantar flexion assistance (Galle et al. 2013a; Malcolm et al. 2013). Therefore, an incremental walking exercise test similar to the weighted walking test (Klimek and Klimek 2007) will be used and the walking performance during this test will be expressed as the weight that subjects are carrying at volitional termination of the test. We choose to focus on the comparison of powered versus unpowered walking because our main research question concerns the influence of push-off assistance during higher intensities. Our first hypothesis (A) is that the assistance of the powered exoskeleton can still reduce metabolic power when compared with an unpowered exoskeleton during walking with high external workloads, induced by a slope and carrying additional weights. As the steering algorithm of the exoskeleton and the air pressure of the pneumatic muscles remain unaltered during the exercise test, we assume that the assistance pattern of the exoskeleton will be similar over increasing weights and will therefore result in an absolute reduction in metabolic power that is similar during the subsequent intervals of the exercise test. Our second hypothesis (B) is that it is possible to reach maximal metabolic effort both with a powered and an unpowered ankle-foot exoskeleton. As a result of these two hypotheses an increase in maximal carried weight and thus walking performance is expected in the powered condition.

3. Methods

Subjects

Nine healthy female subjects [age 21.3 yr (SD 2.2), weight 69.8 kg (SD 9.2), height 171.4 cm (SD 4.6)] participated in the study. Female subjects of normal height and weight were chosen because they fit best in the exoskeleton and because exoskeleton assistance would have a greater effect on their relatively low body weight. They had no previous experience with walking with exoskeletons but all had experience with treadmill walking. All participants gave written informed consent and the protocol was approved by the ethical committee of the Ghent University Hospital.

Exoskeleton

The exoskeleton is a device that can be worn by healthy subjects and that fits around the left and right lower leg (Fig. 1A). It consists of an ankle-foot orthosis with a hinge at the ankle joint and a McKibben pneumatic muscle attached at the dorsal side. The exoskeleton has a weight of 0.76 kg at each foot and fits in sport shoes where footswitches (Multimec 5E/5G, Mec, Ballerup, Denmark) are built in. These footswitches allow to detect foot contact, which is used to impose a specific timing and duration in which the pneumatic muscles are inflated. The pneumatic muscles are connected to air supply and when inflated (air pressure, ± 3.5 Bar) they shorten and cause ankle plantar flexion (Fig. 1B). The goal of our exoskeleton is to add plantar flexion power to the ankle during push-off (Galle et al. 2013a, 2013b; Malcolm et al. 2013). The pneumatic muscles can be turned on and turned off at specific time intervals based on footswitch signals and are triggered with a computer program (Labview, National Instruments, Austin, TX). Start of pneumatic muscle actuation was set at 43 % of stride for level walking (Malcolm et al. 2013) and at 36 % of stride for uphill walking (Galle et al. 2013b) as previous studies showed that these actuation timings are metabolically optimal. Pneumatic muscles were turned off after 63 % of stride, coinciding with toe-off in all conditions. Peak pressure of the pneumatic muscles and peak mechanical power of the exoskeleton occurs in between start and end of pneumatic muscle actuation (Galle et al. 2013a, 2013b; Malcolm et al. 2013).

Protocol

All participants performed two incremental exercise tests, similar to the weighted walking test (Klimek and Klimek 2007) on two different days with one week in between. These tests were performed under two randomized exoskeleton conditions: on one day with actuation of the pneumatic muscles to assist plantar flexion during push-off (powered condition) and on another day without pneumatic muscle actuation (unpowered condition). Before the exercise test subjects performed a standing rest trial of 4 min to determine resting energy expenditure and a 22 min habituation session (Galle et al. 2013a) on a level treadmill (HP Cosmos, Nussdorf-Traunstein, Germany) at 5 km·h^{-1} to learn to walk with the exoskeleton. In the powered condition both the habituation and the exercise test were done with a powered exoskeleton and in the unpowered condition both the habituation and the exercise test were done with an unpowered exoskeleton. The exercise test was performed at 5 km·h^{-1} on a treadmill with a 15 % slope. Subjects walked during 3 min intervals with 1 min of rest in between. In the first interval subjects walked on the treadmill with an unloaded weight vest (no weight) and every 3 min a weight corresponding to 5 % of body weight was added to the weight vest (Fig. 1). Once all compartments were filled (20 kg), a backpack was used to add more weights. In between 3 min intervals, 1 min of rest allowed us to add weights and collect blood lactate samples. Subjects were instructed to continue the walking protocol until voluntarily termination due to exhaustion.

Data Collection

During the entire protocol subjects wore a heart rate belt, a nose-clip and breathed in a mouthpiece. Heart rate (RS 400, Polar, Oulu, Finland), O_2 consumption and CO_2 production (Oxycon Pro, Jaeger GMBH, Höchberg, Germany) were measured during the entire protocol. In the first 30 sec after every 3 min interval 65 µL capillary blood samples were collected from the tip of the middle or third finger of the left hand and analyzed within the next 35 sec with a blood gas analyzer (Radiometer, ABL-90 Flex, Brønshøj, Denmark). At termination of the exercise test subjects were asked to score perceived exertion on the Borg scale (Borg 1973). In 2 subjects this failed due to human errors, resulting in reliable values for 7 subjects. An end-test blood lactate sample was taken 2 min after exercise termination due to weight unloading immediately after exercise termination as subjects were carrying weights of over 20 kg.

Data analysis

End-test blood lactate concentration was the blood lactate concentration at exercise termination and peak heart rate was the highest measured heart rate value, always in the last min of the exercise test. Metabolic energy expenditure was estimated with the formula of Brockway (Brockway 1987) based on 30 sec mean values of O_2 consumption and CO_2 production and divided by subjects body weight to calculate metabolic power (W·kg^{-1}). Metabolic power of the second and third min of the 4 min standing rest trial in the beginning of the experiment was subtracted from gross metabolic power to calculate net metabolic power.

Net metabolic power for all 3 min intervals was calculated based on the last min of each interval. Peak net metabolic power was the metabolic power in the last min of the last completed walking interval. In 2 out of 151 measures, metabolic power was deleted from the analysis as the net metabolic power of the interval was more than 10 % lower compared to the previous interval, which is unlikely and the result of measurement errors (e.g. due to nose clip displacement). $\dot{V}O_2$ peak was determined based on the highest 30 sec mean value of the entire protocol, always in the last min of the exercise test. Maximal carried weight was the weight that subjects were carrying in the last completed 3 min interval. Total weight was defined as the total weight that subjects were moving against gravity, which is the sum of body weight, exoskeleton weight, shoe weight and the additional weight that subjects carried.

As the number of completed intervals varied between subjects, linear regressions were used to model the individual relationship between net metabolic power and the carried weight for each 3 min interval for both the powered and unpowered exercise test. All individual regressions were based on at least 6 data points. The individual y-intercepts and slopes of the regressions could then be averaged to compute the mean linear regression for the powered and the unpowered exercise test. The carried weight in the regression analysis was expressed as a percentage of the maximal weight that subjects carried in the unpowered exercise test, which was referred to as the maximal unpowered weight. Thereby, all subjects terminated the unpowered exercise test with a weight of 100 % of maximal unpowered weight.

Because the number of completed 3 min intervals varied between subjects, it was not possible to calculate population averages for every interval for blood lactate. Therefore, blood lactate concentration was calculated for 3 instants during the unpowered exercise tests: for the first (= without weight), the middle (= 50 % maximal unpowered weight) and the last interval of the exercise test (= maximal unpowered weight). This resulted in 3 lactate values (for the beginning, the middle and the end of the unpowered exercise test) that could be averaged across subjects. These were compared with the lactate averages for the intervals of the powered exercise test with the same carried weight. For an even number of intervals the mean of the two middle intervals was used for the middle value.

A similar approach was used to see if a steady state in the net metabolic power was reached in the last minute of every 3 min interval by comparing the 6 subsequent 30 sec averages of net metabolic power. As the number of completed 3 min intervals varied between subjects, population averages could not be calculated for every interval. Therefore this analysis was done for all subjects for the first (= without weight), the middle (= 50 % maximal unpowered weight) and the last interval of the unpowered exercise test (= maximal unpowered weight). For an even number of intervals the mean of the two middle intervals was used for the middle value. This resulted in net metabolic power for 3 intervals (for the beginning, the middle and the end of the unpowered exercise test) that could be averaged across subjects. These were compared with the intervals of the powered exercise test with the same carried weight.

Exoskeleton mechanical power

During the powered condition, the exoskeleton delivers additional mechanical power to the ankle joint (Galle et al. 2013a; Malcolm et al. 2013). Exoskeleton mechanical power was not measured during the exercise tests but in order to evaluate the magnitude of the exoskeleton assistance, 2 representative subjects out of the 9 subjects performed an additional protocol on a different day. During this protocol, exoskeleton mechanical power was measured for the right leg during uphill walking with progressively higher weight carrying. Subjects walked on an inclined treadmill (15 %) at 5 km·h^{-1} during 1 min intervals with 2 min rest in between. The weight in the different intervals was identical to the weight that these subjects carried in the powered exercise test (0 to 31 kg over 11 intervals) in order to get a reliable estimation for the amount of additional power delivered by the exoskeleton during the consecutive intervals of the previous exercise test. Reflective markers (4) on the shank and foot allowed us to detect foot contact and ankle joint angle with MaxTraq software (Innovasion Systems, Columbiaville, MI, USA) on high speed video recordings (100 Hz, Redlake, Morgan Hill, CA, USA). A load cell (100 Hz, 210 Series, Richmond Industries Ltd., Reading, UK) was connected to the right pneumatic muscle to measure the pneumatic muscle force during walking and was recorded in synchronization with the high speed video during 10 sec, which equals approximately 10 strides of uphill walking. Moment arm of the pneumatic muscle was measured as the perpendicular distance between the exoskeleton joint and the pneumatic muscle for the right leg (=0.08 m). Ankle angle and load cell data were filtered with a 2nd order Butterworth low-pass filter (cut-off frequency 12 Hz). Load cell data were used to calculate pneumatic muscle force and were multiplied with pneumatic muscle moment arm to calculate pneumatic muscle torque of the right leg. Mechanical power of the exoskeleton for the right leg was then calculated by multiplying ankle angular velocity (= first derivative of ankle joint angle) and pneumatic muscle torque and divided by body weight. Exoskeleton power per stride was then averaged to calculate net exoskeleton mechanical power per stride and was averaged for 8 to 9 consecutive strides as previous studies showed rather large variations between steps (Malcolm et al. 2013; Sawicki and Ferris 2008, 2009a).

Statistics

Cohens' d statistics (Cohen 1977) were used to calculate effect sizes for our main outcomes which are the y-intercepts of the linear regressions, indicating a difference in metabolic power between powered and unpowered walking with the same carried weight, and the carried weight in the last interval of the powered and unpowered exercise test, indicating the walking performance in the powered and unpowered exercise test. Effect sizes were higher then 1, indicating large effect sizes.

Regression analysis and other statistics were done with SPSS Statistics 20 (IBM, Armonk, NY). Paired samples t-tests with the α level of significance set at $P \leq 0.05$ were done: to compare end-test physiological parameters and maximal carried weight between the powered and unpowered exercise test; to search for differences in y-intercepts and slopes of the linear regressions of the powered and unpowered exercise test; to compare blood lactate concentration between powered and unpowered walking. Repeated measures ANOVA with post hoc comparisons and Bonferroni correction and with the α level of significance set at $P \leq 0.05$ were done to check for steady state in the 6 subsequent 30 sec averages of net metabolic power of the 3 min intervals in the beginning, the middle and the end of the exercise test.

4. Results

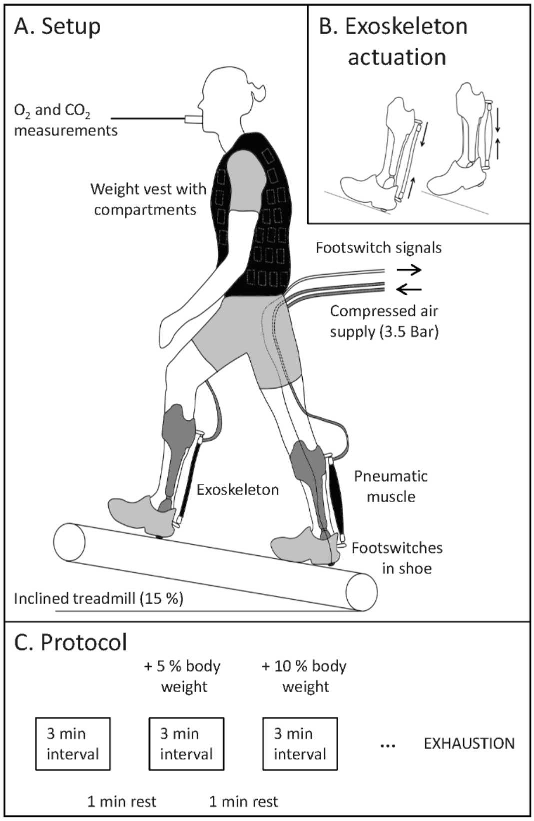

Fig. 1. Experimental setup (A), Exoskeleton actuation (B) and Experimental protocol (C).
(A) Experimental setup: subjects wore an exoskeleton and vest with several compartments that could be filled with weights. During the entire experiment O_2 and CO_2 measurements were recorded with a mouthpiece. In the powered condition footswitches in the heel detected foot contact and the pneumatic muscles of the exoskeleton were turned on during push off by means of compressed air inflation.
(B) Exoskeleton actuation: in the powered condition the inflated air in the pneumatic muscles caused a shortening of the pneumatic muscles and thereby induced plantar flexion assistance during the push-off.
(C) Protocol: subjects performed a powered and unpowered maximal walking exercise test with progressively higher weight carrying on an inclined treadmill (15%) at 5 km·h^{-1}. Every 3 min 5% of body weight was added until subjects terminated the exercise test due to exhaustion. In between the intervals 1 min of rest allowed to add weights and collect blood samples and perception data.

Table 1. End-test physiological and performance measures for the unpowered and powered exercise test

	Condition				*P*
	Unpowered exercise test		Powered exercise test		
End-test blood lactate (mmol· L^{-1})	8.14	(2.24)	7.93	(2.49)	0.811
Peak heart rate (bpm)	191.78	(6.50)	190.00	(6.50)	0.558
End-test Borg scale (6-20 scale)	18.93	(0.73)	18.57	(0.79)	0.182
Peak net metabolic power (W·kg^{-1})	12.52	(0.91)	12.48	(1.02)	0.915
$\dot{V}O_2$ peak (ml·min^{-1}·kg^{-1})	40.55	(3.05)	40.55	(2.78)	0.999
Max. carried weight (kg)	22.47	(3.57)	29.54	(4.13)	≤ 0.001*****

End-test values for the unpowered and powered exercise test. See Methods for calculations. Values are means of subjects (SD) and *P* values are the result of paired samples t-tests to search for differences between the powered and unpowered exercise test. * indicate a significant difference between the unpowered and powered exercise test.

All subjects sustained the powered and unpowered exercise test for at least 6 intervals, thereby carrying more than 30% of body weight during uphill walking when terminating the exercise tests. The end-test physiological parameters that were collected at volitional termination of the exercise test were similar in the powered and unpowered condition: blood lactate concentration, heart rate, Borg scale score, peak net metabolic power and $\dot{V}O_2$ peak did not differ at the end of both exercise tests (Table 1).

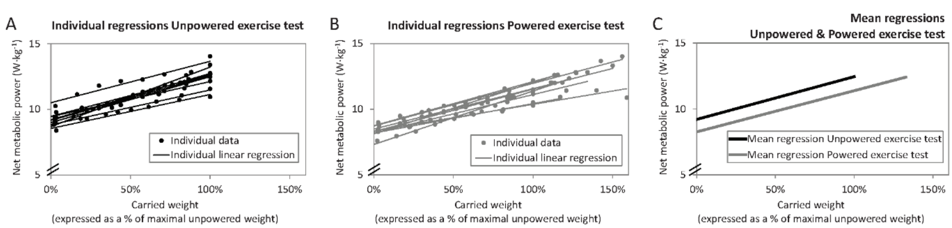

Fig. 2. Linear regression between net metabolic power and carried weight for the exercise tests. All individual data points (●) and individual regressions (thin lines) are plotted for the unpowered (A) and the powered exercise test (B). Based on the individual regressions an average linear regression (thick lines) was calculated for both the powered and unpowered exercise test (C), see Methods and Table 2 for details. Carried weight was expressed as a % of maximal unpowered weight during both exercise tests. In this way the exercise test is terminated with a weight that corresponds to 100% of the maximal unpowered weight for all subjects in the unpowered exercise test.

Subjects increased performance in the powered exercise test compared to the unpowered exercise test (Table 1): subjects were able to sustain the walking protocol longer and thereby carried a 7.07 ± 3.34 kg higher weight at the end of the exercise test in the powered condition. In terms of the total weight that subjects were moving against gravity (which is the sum of body weight, exoskeleton weight, shoe weight and the weight that subjects carried), subjects were able to transport 7.7 ± 4.1 % more total weight against gravity in the powered condition than in the unpowered condition.

Table 2. Linear regression parameters for unpowered and powered exercise test

	Y-intercept of individual regression	Slope of individual regression	Adjusted R^2	*P*
Unpowered exercise test	9.20 (0.58)	3.25 (0.61)	0.97 (0.03)	≤ 0.001° for all
Powered exercise test	8.26 (0.42)	3.13 (0.67)	0.97 (0.04)	≤ 0.001° for all
	0.004*	0.576		

Mean values for the y-intercepts, slopes and adjusted R^2 of the individual linear regressions between net metabolic power versus additional weight expressed as a % of maximum unpowered weight. *P* values on the right refer to linear regression statistics. ° indicate significant linear regression. *P* values at the bottom refer to paired samples t-test for y-intercepts and slopes of unpowered and powered exercise tests. * indicate significant differences between the unpowered and powered exercise test.

In the first interval of the exercise test, when subjects were walking uphill without carrying weights, net metabolic power was 8.0 ± 6.2 % lower for powered exoskeleton walking (8.48 ± 0.42 $W \cdot kg^{-1}$) than for unpowered exoskeleton walking (9.24 ± 0.53 $W \cdot kg^{-1}$). At the end of the exercise test, when carrying a weight corresponding to 100% of the maximal unpowered weight (22.5 ± 3.6kg) net metabolic power was 10.1 ± 6.8 % lower for powered exoskeleton walking (10.85 ± 0.67 $W \cdot kg^{-1}$) compared to unpowered exoskeleton walking (12.52 ± 0.91 $W \cdot kg^{-1}$). A linear regression was done for every individual for net metabolic power versus carried weight, expressed as a % of the maximal unpowered weight (Fig. 2). All individuals showed a significant linear regression between net metabolic power and carried weight for both the powered and unpowered exercise test (Table 2). The y-intercepts of these linear regressions were significantly lower (-0.95 ± 0.72 $W \cdot kg^{-1}$) for the powered exercise tests than for the unpowered exercise tests, indicating lower net metabolic power for walking without weights in the powered condition. No significant difference between the powered and unpowered exercise test could be found for the slopes of the linear regression, indicating a constant difference in net metabolic power. In other words, a parallel linear relationship was found between the powered and unpowered exercise test with a constant significantly lower net metabolic power over the entire protocol in the powered condition.

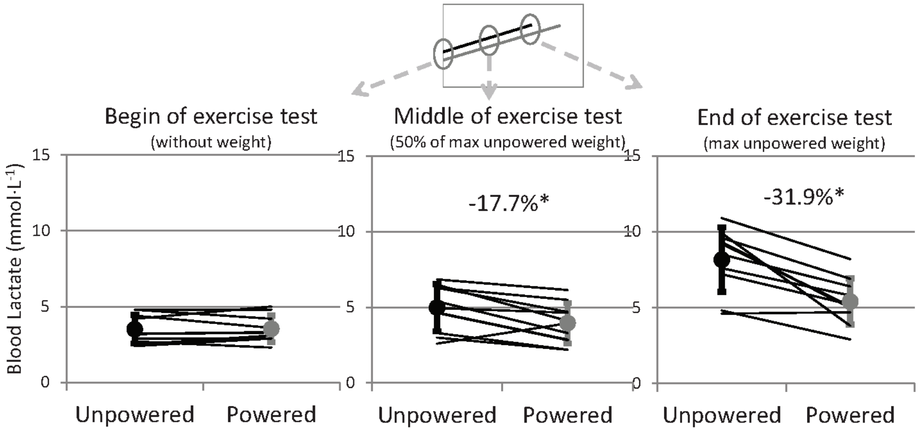

Fig. 3. Blood lactate concentrations for different weight conditions. Mean (●) and individual (thin lines) capillary blood lactate concentrations (mmol·L^{-1}) are compared between the powered and unpowered exercise test in the beginning (walking without weights), the middle (with carrying 50% of the maximal unpowered weight =11.5 ± 1.7 kg) and at the end of the exercise tests (with carrying 100% of the maximal unpowered weight =22.5 ± 3.6 kg). Error bars are ± 1 SD of the mean. * indicate significant difference between the powered and unpowered exercise test with paired samples t-test ($P \leq 0.05$). Percentages express the difference in capillary blood lactate concentration between the unpowered and powered exercise test.

The exoskeleton mechanical power measurements that were collected in a subsample on an additional test day showed only small variations over increasing weights (0 to 31 kg). Net exoskeleton mechanical power per stride for the right leg was 0.13 ± 0.01 W·kg^{-1} and 0.12 ± 0.01 W·kg^{-1} for the different intervals with increasing weight for respectively the first and second subject.

In the first interval of the exercise test, when subjects were not carrying weights, blood lactate concentration was not significantly different between the powered condition (3.54 ± 0.87 mmol·L^{-1}) and the unpowered condition (3.51 ± 0.95 mmol·L^{-1}) when walking without weights. During increasing intensities, blood lactate concentration was significantly lower in the powered condition (3.96 ± 1.34 mmol·L^{-1}; 5.40 ± 1.52 mmol·L^{-1}) compared with the unpowered condition (4.99 ± 1.56 mmol·L^{-1}; 8.16 ± 2.12 mmol·L^{-1}) in the middle and at the end of the exercise test, when walking with a weight corresponding to respectively 50 % (11.5 ± 1.7 kg) and 100 % (22.5 ± 3.6 kg) of the maximal unpowered weight (Fig. 3).

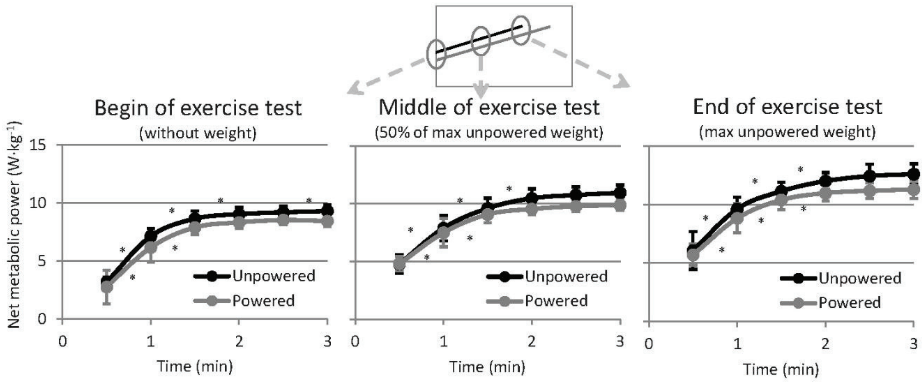

Fig. 4. Changes in net metabolic power during 3 min intervals. The subsequent 30 sec averages for net metabolic power during a 3 min interval are averaged across subjects in the beginning (walking without weights), the middle (with carrying 50% of the maximal unpowered weight =11.5 ± 1.7 kg) and the end of the exercise tests (with carrying 100% of the maximal unpowered weight =22.5 ± 3.6 kg). * indicate significant differences between consecutive 30 sec averages in net metabolic power with Repeated measures ANOVA with post hoc comparisons and Bonferroni correction (*P* ≤ 0.05).

Analysis of the net metabolic power for consecutive 30 sec averages within a 3 min interval in the beginning, the middle and at the end of the exercise test indicated a steady state in net metabolic power after 2 min as no significant differences were found between consecutive 30 sec averages after 2 min (Fig. 4). In the first unpowered condition a significant difference was found between min 2.5 and min 3 but the difference of 0.12±0.08 W·kg^{-1} seems negligible.

5. Discussion

The aim of our study was to quantify walking performance with a simple powered exoskeleton during a maximal exercise test. During the exercise test, assistance of the exoskeleton resulted in a reduction in net metabolic power of 8 % at lower intensities (walking without weights) and 10% at higher intensities (walking with 100% of maximal unpowered weight). This percentage reduction in net metabolic power is similar with previous research during assisted loaded walking (Mooney et al., 2014) or uphill walking without weights (Sawicki and Ferris 2009a) and emphasizes that subjects can benefit from plantar flexion assistance during uphill loaded walking. Also the lactate values that did not differ between both conditions in the beginning of the exercise test and that differed when intensities increased indicate that the lactate threshold is exceeded later in the exercise test in the powered condition and shows that plantar flexion assistance reduces the effort for a specific workload. These findings are consistent with the regression analysis that showed a parallel linear relationship for powered and unpowered walking when net metabolic power was plotted against carried weight, with a constant reduction of 0.95 ± 0.72 $W \cdot kg^{-1}$ in the powered condition. This confirms our first hypothesis (A): powered ankle-foot exoskeletons can reduce the metabolic power when compared with unpowered walking, also during maximal exercise intensities.

At volitional termination of the exercise tests, end-test physiological measures ($\dot{V}O_2$ peak, peak net metabolic power, peak heart rate, blood lactate concentration and Borg score) were similar for the powered and unpowered condition. While these end-test measures are a little lower than expected for a maximal exercise test, they are corresponding to the reported values for the weighted walking test (Klimek and Klimek 2007) and situated within the boundaries of reported maxima for exercise testing (Herdy and Uhlendorf 2011; Koch et al. 2009; Midgley et al. 2007). Therefore, given the specific characteristics of this uphill walking task, subjects reached maximal oxidative metabolism and at least a near maximal effort. As the effort at volitional termination of the powered and unpowered exercise test can be considered similar and (near) maximal, this confirms our second hypothesis (B) that it is possible to reach maximal metabolic effort both with a powered and an unpowered ankle-foot exoskeleton.

The 3 min intervals that were used in our exercise test might be too short to reach a true steady state in net metabolic power, given the possible slow component in $\dot{V}O_2$ at higher intensities (Zoladz and Korzeniewski 2001). However, analysis of the subsequent 30 sec averages indicate a steady state in the last minute of the interval and intervals with the same duration were also used in a similar exercise test (Klimek and Klimek 2007). As it was not our aim to measure a true $\dot{V}O_2Max$ but to compare the powered and unpowered condition, it seems unlikely that longer intervals would have altered our main conclusions.

Because of the constant reduction in net metabolic power when walking with a powered exoskeleton and because subjects reached the same (near) maximal effort, subjects were able to increase their walking performance by carrying 7.07 ± 3.34 kg more additional weight. When body weight, exoskeleton weight, shoe weight and additional carried weight are taken into account, 8% more total weight was transported against gravity during uphill walking with a powered exoskeleton. Although a training effect might influence our results in the subjects that performed the unpowered exercise test first, all subjects showed an increase in walking performance in the powered condition. We are not aware of any other scientific reports on successful increases in performance during locomotion with the use of exoskeletons. The ratio of the increase in walking performance corresponds to the reduction in metabolic power of 8 to 10% for powered walking versus unpowered walking with the same weight. This emphasizes the direct relationship between the reduction in metabolic power with a powered exoskeleton and the resulting increase in weight carrying performance.

Exoskeleton mechanical power measurements in a subsample of 2 subjects indicate that net exoskeleton mechanical power per stride did not change over increasing weights, which seems in agreement with the constant reduction in net metabolic power based on the regression analysis. Although we did not measure exoskeleton mechanical power during the exercise test, the values of ~0.12 and ~0.13 $W \cdot kg^{-1}$ per leg that were collected on a separate day are in line with the results of other studies on exoskeleton walking (Malcolm et al. 2013; Mooney et al. 2014; Sawicki and Ferris 2008, 2009a, 2009b). The muscular efficieny of positive joint power during steep uphill locomotion is assumed to be ~0.25 (Margaria 1976), which means that the addition of 1 W in positive mechanical exoskeleton power can be expected to cause a reduction in metabolic power of 4 W. It is therefore not surprising that ~0.25 $W \cdot kg^{-1}$ (for both legs) of exoskeleton mechanical power results in a constant reduction of 0.95 ± 0.72 $W \cdot kg^{-1}$ in net metabolic power. Estimations based on the literature (Lay et al. 2007; McIntosh et al. 2006) indicate that the amount of net exoskeleton power that was added to the ankle joint during the powered exercise test represents around 30 % of normal net ankle joint power during uphill walking. Much attention is paid to loaded walking in a military context as this can determine mission success (Knapik et al. 2012). Developing an exoskeleton that allows to carry more load, increase endurance or reduce metabolic power seems not straightforward (Zoss et al. 2006). Of the few carefully controlled scientific reports of exoskeletons that are intended to increase performance or allow to perform tasks with lower metabolic power, most of them point in the direction of an increase in metabolic power or a decrease in performance when walking with these devices (Gregorczyk et al. 2006, 2010, 2012; Kazerooni and Steger 2006; Pratt 2004; Schiffman et al. 2010; Walsh et al. 2007). Current commercial exoskeletons intended for load bearing are based on structures that transfer the load via rigid beams towards the ground, which reduces the stress on the human musculoskeletal system but increases the mass of the device. Mooney et al. (2014) showed that it is possible to reduce metabolic power of walking with weights with an exoskeleton that has no such load transferring structure but solely relies on assistive power that acts distally at the ankle. While this has the potential disadvantage that increasing the load also increases the stress on the human musculoskeletal system, it reduces the mass of the device. Our results indicate that plantar flexion assistance can increase the weight that subjects can transport with 7.07 ± 3.34 kg and that walking endurance can be increased as subjects continued the exercise test longer. This suggests that also more simple approaches can be useful to assist loaded walking.

In general, our results emphasize the potential of acting on the ankle joint in assistive devices which could be applied in clinical rehabilitation exoskeletons as the ankle-foot complex is often not incorporated (Duerinck et al. 2012), in healthy people to increase performance in terms of endurance and strength and in experimental studies to answer fundamental questions. As subjects were able to use the assistance of the exoskeleton even when fatigued and when carrying heavy loads, future exoskeleton studies can be done in more challenging experimental settings.

The major limitation towards an implementation of our exoskeleton in daily life applications is that it is not autonomous but Mooney et al. (2014) showed that with an altered design, plantar flexion assisting exoskeletons can be made autonomous . However, we see a role for biomechanists and physiologists in studying the human-exoskeleton interaction without being concerned about technical or practical limitations. While this allows to study more fundamental topics (Sawicki and Ferris 2008, 2009a, 2009b), specific knowledge that results from these studies, e.g. on optimal actuation timing (Malcolm et al. 2013), can also be used in the development of autonomous exoskeletons (Mooney et al. 2014). Our choice to compare powered and unpowered walking can also be seen from this perspective as it allows to specifically study the effect of the power assistance to the ankle isolated from the possible side effects of wearing the passive structure of the exoskeleton. Towards practical relevance, the comparison with a standard shoe condition, which is missing in our study, would allow to study the increase in metabolic power because of the movement restrictions and the supplementary weight of the exoskeleton.

We are aware of the smaller reductions in net metabolic power compared to the results with a similar exoskeleton during level walking (Malcolm et al. 2013) and also previous results of uphill walking with our exoskeleton revealed larger reductions in net metabolic power (Galle et al., 2013b). In order to allow the pneumatic muscles to function continuously during more than 30 min, additional air filters were added to the hardware to prevent dust and water vapor to enter the pneumatic muscles. Post-experiment analysis showed that this modified both the magnitude and the timing of the assistive power. As a previous study showed that actuation timing is an important determinant of the reduction in metabolic power (Malcolm et al., 2013), timing changes may have reduced the effectiveness of the exoskeleton during the exercise test.

In conclusion, we studied walking performance of a powered ankle-foot exoskeleton, which seems relevant for military, recreational and experimental purposes, during a maximal uphill walking exercise test with increasing weights up to 30kg. Assistance of a powered exoskeleton reduced the metabolic power of walking by more than 8% for each weight and allowed subjects to continue the exercise test longer, thereby carrying 7.07±3.34 kg more weight. Although our exoskeleton is not autonomous and a comparison with walking with standard shoes is missing, the results advance exoskeletal research and development into a new level: increasing walking performance during maximal exercise by increasing maximal carried weight during an exercise test with progressively higher weight carrying. We demonstrated that it is possible to reach (near) maximal effort during exoskeleton walking and that plantar flexion assistance is still effective during higher intensities. Our results emphasize the potential of simple ankle-foot exoskeletons and the importance of acting on the ankle joint in assistive devices, even during more challenging tasks .

Acknowledgements

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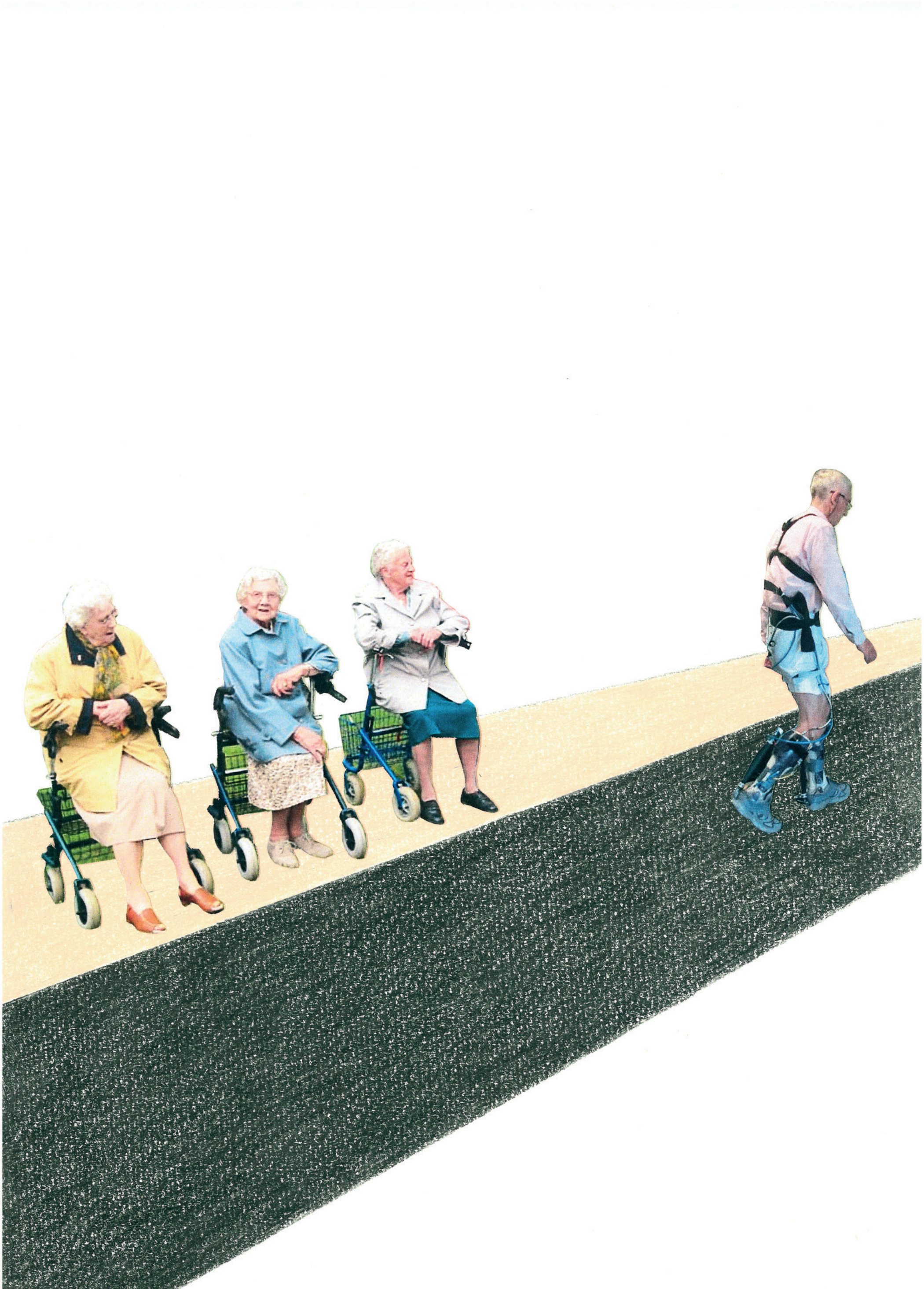
CHAPTER 6



WALKING WITH A PLANTAR FLEXION ASSISTING EXOSKELETON IN HEALTHY ELDERLY

Galle S, Derave W, Bossuyt F, Calders P, Malcolm P, De Clercq D. (Submitted Gait & Posture may 2015)

1. Abstract

Elderly are confronted with reduced physical capabilities and increased energy cost of walking. Exoskeletons that assist walking have the potential to reverse these trends. The goal of the present experiment was to study the feasibility and the effect of a plantar flexion assisting exoskeleton on the metabolic energy cost of walking in physically active and healthy elderly. Seven elderly (age 69.3 ± 3.5 y) walked on treadmill (1.11 $m{\cdot}s^2$) with normal shoes and with the exoskeleton both powered and unpowered. After 20 min of habituation, subjects were able to walk with the exoskeleton without falling or stumbling and assistance of the exoskeleton resulted in a reduction in metabolic cost of 12 % versus unpowered exoskeleton walking. Also, walking with the exoskeleton was perceived less fatiguing for the muscles compared to normal walking, while balance was not perceived worse compared to normal walking. No significant reduction in metabolic cost versus walking with normal shoes was found due to the penalty of wearing the unpowered exoskeleton and the relatively low exoskeleton mechanical power. With improvements in design and actuation of the exoskeleton, it is expected that reductions in metabolic cost compared to walking with normal shoes are possible in healthy elderly, which can have health benefits to counter reduced exercise tolerance. We showed that healthy elderly can walk with an exoskeleton and that this resulted in a reduction in metabolic cost compared to unpowered walking, which is promising for future exoskeleton research that should focus on populations with reduced exercise tolerance or reduced physical function that can benefit from exoskeleton walking assistance.

2. Introduction

Plantar flexion assisting exoskeletons can reduce the metabolic energy cost of walking, increase performance or preferred walking speed in healthy subjects [1–5], or restore function in impaired subjects [6]. With the development of autonomous exoskeletons, which reduce metabolic energy cost during walking without external energy input by exploring passive mechanisms [3] or while carrying all the hardware and power sources [2,7], mobility assistance in daily life becomes a realistic option. Exoskeleton experiments with healthy subjects mainly have applications in a military setting, e.g. for load carrying. The logical next step in exoskeleton research is to focus on specific populations in which the implementation of an exoskeleton can improve quality of life, e.g. in stroke patients [6].

Another valuable target population are elderly, in specific those with reduced exercise tolerance. While metabolic energy cost of walking increases [8], maximal aerobic power and physical capabilities reduce with age [9,10], which may lead to limited performance in daily life activities. As the resulting small aerobic reserves during walking are related to the reduced walking speed in elderly [11], a reduction in the metabolic energy cost of walking with an exoskeleton could allow increased performance [4], while still maintaining a comfortable aerobic reserve. Also, low cardiorespiratory fitness is a predictor for mortality risk in elderly [12], but performing physical activity can improve cardiorespiratory fitness, also in sedentary elderly [13], and reduce mortality risk [14]. However, reduced physical capabilities and walking mobility can limit the possibilities to be physically active. Walking assistance with an exoskeleton could thereby, apart from increases in social participation and quality of life, also improve walking mobility and physical activity, similar to the assumed positive effect of electrical bicycles on meeting physical activity guidelines for sedentary people [15]. Exoskeletons could then help the elderly to keep on walking and stay one step ahead of the Grim Reaper [16].

It is important to first focus on healthy elderly that experience normal age-related biological degeneration before exoskeleton assistance can be applied in elderly with pathologies. While our exoskeleton reduces metabolic cost up to 10% [17] versus normal walking in a young population, it is unclear how elderly will perform with our exoskeleton, due to reduced function [9,10] and alterations in gait [18]. Gait of elderly is characterized by reduced walking speeds and reduced step length [18], which may be related to plantar flexor weakness [19] and were found to have a negative effect on exoskeleton performance [20]. Also, stability is a concern during exoskeleton locomotion [21] and elderly show reduced dynamic balance [18]. However, the exoskeleton specifically assists the weakened plantar flexors, which is an additional argument for a plantar flexion assisting exoskeleton in elderly and Norris et al. [5] showed that elderly can walk with a plantar flexion assisting exoskeleton although they did not find a significant increase in preferred walking speed or a reduction in metabolic cost.

Our goal is to study the feasibility and the efficacy of exoskeleton assistance, focused on metabolic cost, during walking with a plantar flexion assisting exoskeleton in healthy elderly (age 65 or more). Comparison with younger subjects, measurements of basic kinematics, spatiotemporal factors and perception will be used to clarify potential limitations of exoskeleton assistance in elderly.

3. Methods

Subjects

Eight physically active elderly which had no experience with treadmill nor exoskeleton walking signed an informed consent, approved by the ethical committee of the Ghent University Hospital and completed the short self-administered International Physical Activity Questionnaire (IPAQ) [22]. One subject was excluded from the analysis because he was unable to comfortably walk with the exoskeleton due to different walking kinematics (see Supplementary for more details), resulting in an experimental population of 7 subjects (6 male, 1 female; age 69.3 ± 3.5 y; body mass 73.1 ± 6.9 kg; stature 170.4 ± 6.2 cm; European shoe size 42.1 ± 1.8). Reference data of 9 younger adults [17] are added for illustrative purpose in Supplementary.

Exoskeleton

Subjects walked with bilateral ankle-foot exoskeletons that assist plantar flexion during the push-off. More details can be found in previous publications [1,4,23]. Actuation onset was set at 49±1% of the stride and ending was coinciding with toe-off (~68% of the stride). Air pressure was set at 2 bar so that average positive exoskeleton ankle joint mechanical power during a stride equaled 0.11±0.2 $W \cdot kg^{-1}$ for the right leg. It was chosen to focus on 'mild' exoskeleton assistance in the elderly instead of the metabolically optimal actuation onset of ~43% and average positive mechanical exoskeleton power (sum of both legs) of ~0.4 $W \cdot kg^{-1}$ [17] as previous experiments indicated that early actuation timings and high amounts of mechanical power might feel less comfortable.

Experimental design

Subjects performed two walking protocols on treadmill on two separate days with maximum one week in between. First they performed 6 trials of 5 min with 3 min of rest in between: walking with normal shoes in which speed gradually increased from 0.83 $m \cdot s^{-1}$ to 1.11 $m \cdot s^{-1}$ (which was the treadmill speed for all other conditions), walking with the unpowered exoskeleton and four trials walking with the powered exoskeleton to allow total habituation time of 20 min [23]. On the second day, subjects did a 4 min standing rest trial, followed by four walking trials of 5 min with 3 min of rest in between in semi-randomized order (the shoe condition was always first or the last due to the required time to don and doff the exoskeleton). Subjects walked with normal shoes (SHOES), with the unpowered exoskeleton (UNPOW) and twice with the powered exoskeleton (referred to as POW1 for the first powered condition that was done and POW2 for the second powered condition that was done). It was chosen to do the powered condition twice as previous analysis showed that metabolic cost can be increased in the first minutes of powered locomotion because habituation from the previous day is not fully retained. Due to the risk of cumulated fatigue, not all conditions were done twice.

Data collection

All measurements were done on the second day. O_2 consumption and CO_2 production were measured continuously (K4b^2, Cosmed, Rome, Italy). A linear displacement sensor (100 Hz; SLS130, Penny&Giles, Christchurch, United Kingdom) was connected between the right foot and shank segment of the exoskeleton. Based on a 3D calibration with motion analysis, ankle joint angle and moment arm of the pneumatic muscle force of the right leg were estimated using the linear displacement sensor. A load cell (100 Hz; 210 Series, Richmond Industries Ltd., Reading, United Kingdom) was attached to the right pneumatic muscle to measure pneumatic muscle force and consumer camera's (30 Hz; HDR-CX240E, Sony, Weybridge, United Kingdom) collected dorsal images.

After each trial, subjects scored perception on a visual analog scale of 100 mm by setting a mark on a horizontal line with a positive and negative option on each side and a neutral statement in the middle [24]. Subjects scored difficultness of normal treadmill walking (versus normal over ground walking) and difficultness of powered and unpowered exoskeleton walking on treadmill (versus normal walking on treadmill) between 'much more difficult' and 'much easier'. Powered exoskeleton walking was compared with normal treadmill walking for pleasantness from 'much less pleasant' to 'much more pleasant', overall fatigue and leg muscle fatigue from 'much less tiring' to 'much more tiring' and balance from 'much worse' to 'much better'.

Data analysis

IPAQ was used to classify activity level as low, moderate or high [22]. Perception scores were measured as a positive or negative distance from the neutral statement in mm. Net metabolic power was calculated with the formula of Brockway for the last 2 min of each interval, divided by bodyweight and diminished with metabolic power of the standing rest trial (e.g. [1,23]). MaxTraq software (Innovasion Systems, Columbiaville, Michigan, USA) was used to analyze video images to calculate step length (time between consecutive heel contacts multiplied with 1.11 $m \cdot s^{-1}$) and step width (distance between the middle of the left and right heel during stance [25]). Step length and step width were averaged over the last 2 min of each condition and intra-subject variability was calculated as the standard deviation of consecutive steps in the last 2 min of each condition. Analog data were filtered with a 2nd order low pass filter (cut-off 12 Hz) in Matlab (MathWorks Inc., Natick, Massachusetts, USA). Data of the linear displacement sensor were used to calculate ankle joint angle, ankle joint angular velocity and moment arm of the pneumatic muscle force. Exoskeleton torque for the right leg was calculated by multiplying pneumatic muscle force (divide by body mass) with moment arm length and exoskeleton mechanical power for the right leg was calculated by multiplying exoskeleton torque with ankle joint angular velocity. These data were averaged for the last 2 min of the exoskeleton conditions and time-normalized from heel contact to heel contact for the right leg. Average positive exoskeleton mechanical power was averaged for the right stride.

Statistics

All statistics were done with SPSS Statistics 21 (IBM, Armonk, NY, USA). Repeated measures ANOVA ($p \leq 0.05$) with post-hoc tests were done to compare metabolic power, step length, step width, intra-subject variability (on step-length and step-width), average positive exoskeleton power, peak power and perception between conditions. One-sample t-test ($p \leq 0.05$) was done to analyze if perception scores differed from 0.

4. Results

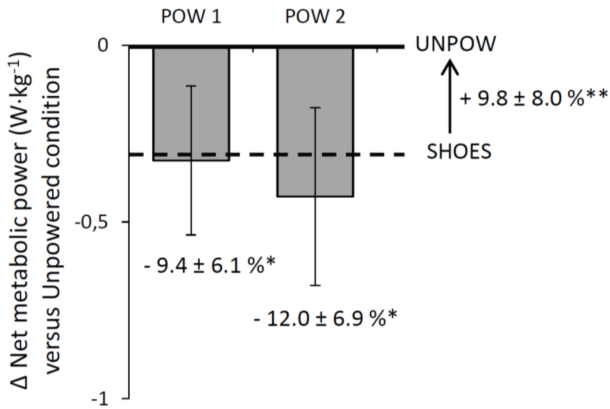


Fig. 1. Net metabolic power. The differences in net metabolic power versus the unpowered condition (UNPOW) are shown for both powered conditions (± s.d) (POW 1 and POW 2). Percentages under the bars represent the reduction for POW1 and POW2 versus UNPOW. The straight line indicates the net metabolic power for UNPOW and the dashed line indicates the net metabolic power for walking with standard shoes (SHOES). Repeated measures ANOVA was done to compare between conditions. * indicate a sign. difference with UNPOW, ** indicate a sign. difference with SHOES ($p \leq 0.05$).

Physical activity level was moderate or high for all subjects. They were all able to walk on the treadmill, with and without exoskeleton, without falling or stumbling. One subject was excluded from the analysis because of walking kinematics that deviated from a 'normal' walking pattern and that disturbed the human-exoskeleton interaction (Supplementary).

Net metabolic power in the powered conditions (POW 1: 3.23±0.49 W·kg^{-1}; POW 2: 3.12±0.40 W·kg^{-1}) was significantly lower than UNPOW (3.55±0.38 W·kg^{-1}) but the weight of the exoskeleton caused a significant increase in UNPOW compared to SHOES (3.24±0.36 W·kg^{-1}) (Fig. 1). Despite the large standard deviation, due to individual differences and the variability of the metabolic measurements during slow walking, net metabolic power was lower for all subjects in POW2 compared to UNPOW, indicating that all subjects were able to benefit from plantar flexion assistance after a short habituation. During powered exoskeleton walking the pneumatic muscles caused an ankle plantar flexion torque and the exoskeleton delivered plantar flexion power (Fig. 2). Ankle angular velocity, exoskeleton torque and exoskeleton mechanical power were very similar for POW1 and POW2 and differed from UNPOW during the push-off (Fig. 2). The exoskeleton delivered 0.11±0.02 W·kg^{-1} average positive mechanical power for the right leg during both POW1 and POW2 (Table 1). In a previous experiment, similar exoskeleton mechanical power assistance resulted in reductions in net metabolic power of 16% (significant) and 5%

(not significant) for powered walking versus respectively unpowered walking and walking with normal shoes in 9 healthy subjects (Supplementary).

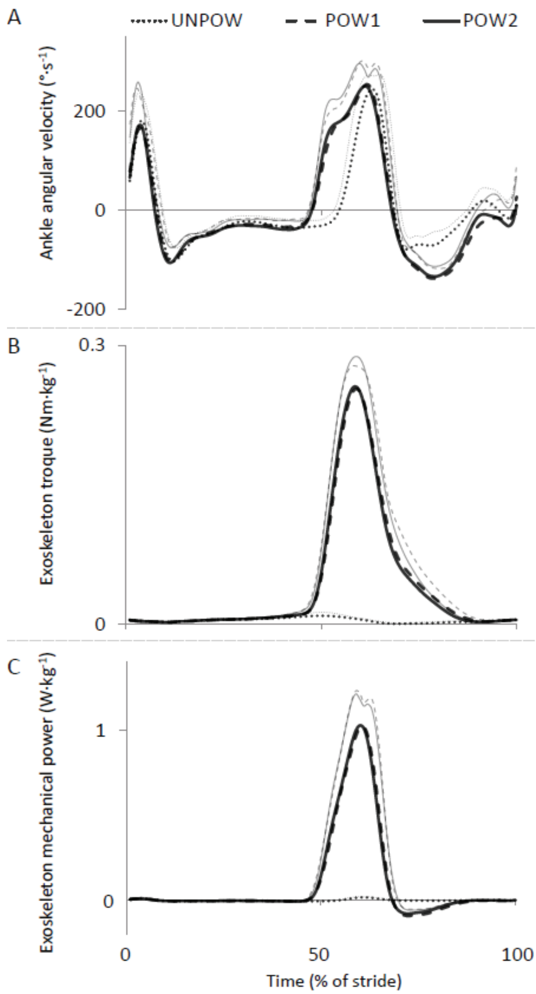


Fig. 2. Joint kinematics and exoskeleton kinetics. Thick lines are population means for POW1, POW2 and UNPOW of ankle angular velocity (A), exoskeleton ankle joint torque (B) and exoskeleton mechanical power (C). Thin curves are + 1 s.d. Curves are normalized for the right leg from heel strike (0%) to heel strike (100%).

Exoskeleton assistance increased step length in POW1 and POW2 compared to UNPOW and increased step length in POW2 compared to SHOES (Table 1). Step width and intra-subject variability for step length and step width showed no significant differences between conditions. Perception of difficulty for walking on treadmill with normal shoes was not significantly different from over ground walking (-

0.29±0.95; p=0.457). During walking on treadmill, UNPOW was perceived more difficult compared to SHOES and assistance of the exoskeleton in POW2 was perceived to cause less fatigue in the legs compared to SHOES (Fig. 2). Although no statistical differences were found between POW1 and POW2 for separate variables, on average perception of all variables was better in POW2.

Table 1. Exoskeleton kinetics and temporal parameters

	SHOES	UNPOW	POW1	POW2
Peak exoskeleton mech. power ($W \cdot kg^{-1}$)	/	0.02 ± 0.00	1.13 ± 0.16*****	1.13 ± 0.16*****
Average pos. exoskeleton mech. power per stride for right leg ($W \cdot kg^{-1}$)	/	0.00 ± 0.00	0.11 ± 0.02*****	0.11 ± 0.02*****
Step length (m)	0.59 ± 0.04	0.59 ± 0.03	0.61 ± 0.05*****	0.62 ± 0.05***,****
Step length variability (m)	0.04 ± 0.01	0.04 ± 0.01	0.04 ± 0.01	0.04 ± 0.01
Step width (m)	0.10 ± 0.02	0.10 ± 0.01	0.10 ± 0.02	0.09 ± 0.01
Step width variability (m)	0.02 ± 0.00	0.02 ± 0.00	0.02 ± 0.01	0.02 ± 0.01

Mean values for mechanical exoskeleton power and temporal parameters during walking (see 'Methods' for calculation). Peak and average exoskeleton mechanical power are based on the mechanical power of the exoskeleton of the right leg (Fig 2C). Temporal parameters are the mean values for step length and step width for the last two minutes of each condition. Step length and step width variability are the standard deviation of the consecutive steps in the last two minutes of each condition for respectively step length and step width. Repeated measures ANOVA was done to compare between conditions. * indicate sign. different from UNPOW, ** indicate sign. different from SHOES ($p \leq 0.05$).

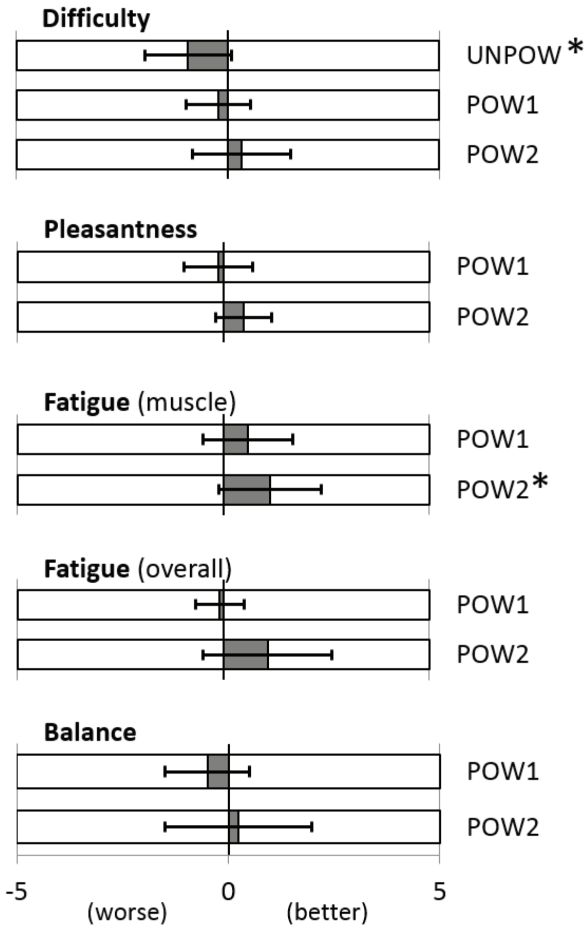


Fig. 3. Perception measures. Horizontal bars show the perception of difficulty for UNPOW, POW1 and POW2 compared to walking with normal shoes on treadmill (SHOES) and the perception of pleasantness, fatigue (for the muscles), fatigue (overall) and balance for POW1 and POW2 compared to SHOES. Subjects had to draw a line on a continuous scale between a negative and a positive option with a neutral statement in between (see 'Methods' for calculation). A negative score represents 'worse' compared to SHOES, a positive score represents 'better' compared to SHOES. A score of 0 represents similar to SHOES. Independent samples t-test was done to compare the perception with the neutral statement (score = 0) and thus compared with SHOES. * indicate sign. different from 0 ($p \leq 0.05$).

5. Supplementary data

During the experiment, one subject (male; age 70 y; body mass 60.0 kg; stature 174.0 cm; European shoe size 40.0) walked with knee flexion during the entire stride, both with and without the exoskeleton (UNPOW_EXCL and POW2_EXCL). Due to knee flexion at heel contact and during the stance phase, lower limb configuration and ankle angle were changed, which resulted in reduced exoskeleton function. While the subject was able to walk with the exoskeleton it was clear that the human-exoskeleton interaction was not optimal and the subject reported discomfort. Due to increased ankle dorsiflexion, the exoskeleton caused plantar flexion force earlier in the stance and higher peak forces, even during unpowered exoskeleton walking. The exoskeleton torque pattern differed strongly from the mean of the other subjects (Fig S1B), which resulted in inefficient exoskeleton assistance and an increase in net metabolic cost of more than 20% for unpowered walking and more than 30% for powered walking compared to walking with normal shoes. As this subject was not representative for our population of healthy and active elderly without gait impairments, this subject was excluded from the analysis.

In a previous experiment (Galle, Malcolm, Collins, et al., 2015), 9 subjects (female; age 21.7 ± 1.0 y; body mass 60.2 ± 4.1 kg; stature 169.3 ± 4.3 cm; European shoe size 38.8 ± 0.8) walked in several exoskeleton conditions from which 1 resulted in a similar exoskeleton mechanical power curve compared to the experiment in elderly (Fig S1C). While angular velocity was higher during the push-off compared to our older subjects, pneumatic muscle force was lower, which resulted in similar amount of exoskeleton mechanical power over a stride. In these younger subjects, assistance of the exoskeleton resulted in a significant reduction in metabolic cost of 16% versus unpowered walking and not significant reduction of 5% versus walking with standard shoes. While the differences between the results in the older population and our previous results in a younger population could be the result of a different experimental protocol, different walking speeds (1.25 $m \cdot s^{-1}$ instead of 1.11 $m \cdot s^{-1}$), differences in habituation duration, different actuation behaviour (e.g. due to length of the pneumatic muscles), or differences in how these populations cope with exoskeleton assistance, it shows that the reductions in metabolic cost because of walking with a similar amount of exoskeleton mechanical power are more or less similar in a younger and an older population where a significant reduction on 12% was found versus unpowered walking and a not significant reduction of 4% versus walking with normal shoes.

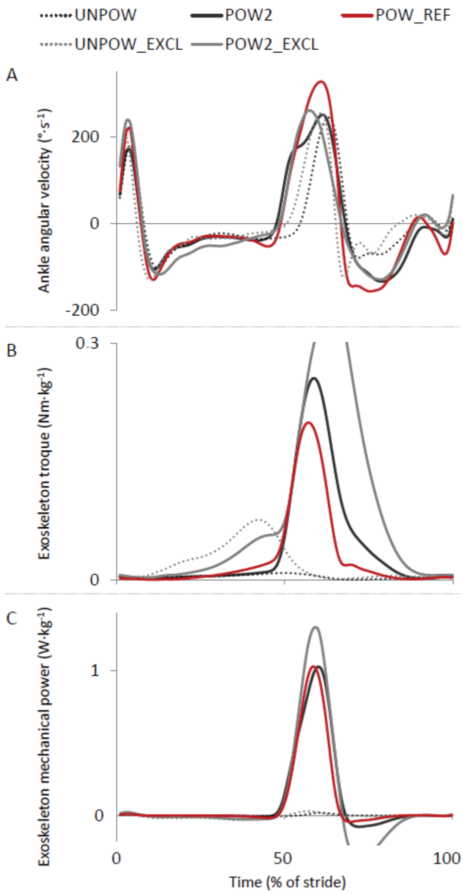


Fig. S1. Joint kinematics and exoskeleton kinetics. Lines are population means for UNPOW and POW2 from Fig. 1, supplemented with the mean of the subject that was excluded (UNPOW_EXCL and POW_EXCL) from the analysis due to altered gait kinematics and reference data of 9 younger subjects (POW_REF) that walked with a similar exoskeleton assistance in a previous experiment (Galle, Malcolm, Collins, et al., 2015). Data are ankle angular velocity (A), exoskeleton ankle joint torque (B) and exoskeleton mechanical power (C), normalized for the right leg from heel strike (0%) to heel strike (100%). No statistical comparison was done as the goal of this comparison is rather illustrative.

6. Discussion

The goal of this study was to investigate the effect of walking with an exoskeleton in an older population. Although one subject was excluded from the analysis due to walking difficulties, all subjects were able to walk with the exoskeleton. Assistance of the exoskeleton, which was perceived as less fatiguing for their muscles compared to normal walking, resulted in a reduction in metabolic cost of 12% versus unpowered walking after a habituation of only 20 min. However, no significant differences were found in metabolic cost between powered walking and walking with normal shoes due to the penalty of 10% caused by the weight of the exoskeleton.

In another study, exoskeleton walking in elderly did not result in a reduction in metabolic cost of transport [5]. Despite difference in habituation, exoskeleton design and control, two main factors seem related to the discrepancy with our results. First, the peak mechanical power of the exoskeleton (0.46 ± 0.12 $W \cdot kg^{-1}$), and hence the average mechanical power, was much lower compared to our results and the amount of exoskeleton power is crucial in reducing the metabolic cost [17]. Our results show that healthy elderly can walk with more exoskeleton power, which results in a reduction in metabolic cost versus unpowered exoskeleton walking. Another factor explaining our good results could be the physical capabilities of our experimental population. Physical activity level was moderate or high in all subjects, which was emphasized in their ability to walk comfortably on the treadmill within 5 min while normal habituation to treadmill walking takes longer [26]. Also, all subjects were able to walk with the exoskeleton without holding the treadmill. Future research with exoskeletons in elderly or patient populations should include physical activity and physical fitness as possible confounding factors.

Our results were compared with unpublished data of 9 younger subjects that walked with the same exoskeleton with similar exoskeleton assistance (Supplementary). Despite differences in protocol, walking speed and habituation, the average (significant) reduction in metabolic cost of 16% versus unpowered walking and (not significant) reduction of 5% versus walking with normal shoes do not strongly differ from our older population (resp. reductions of 12% and 4%). It seemed that the results in the elderly, which could be related to their good physical health, do not strongly differ from the results of a younger population. In this younger population, doubling the amount of exoskeleton power caused significant reductions in metabolic cost of more than 20% versus unpowered walking and 10% versus walking with normal shoes [17]. As we chose to use 'mild' exoskeleton assistance in the elderly instead of the optimal assistance parameters [1,17], it is not surprising that we were unable to find a significant reduction versus normal walking. Overall, elderly did not perceive the actual exoskeleton assistance as disturbing their balance, suggesting that exoskeleton walking with more exoskeleton power is possible, which could lead to higher reductions in metabolic cost. However, caution is necessary with the results of our perception measures with visual analog scales [31]. Also, reducing the weight and the design of the exoskeleton can reduce the penalty of wearing the exoskeleton to app. 5 % instead of 10%, even for autonomous exoskeletons [2,3]. This suggests that reductions of more than 10% versus walking with normal shoes are possible in healthy and physically active elderly by applying more exoskeleton power and improving the weight and the design of the exoskeleton. Another argument towards more exoskeleton power is the plantar flexor weakness and the reduced ankle power generation in elderly [18,19]. It is possible that elderly could thereby benefit more from plantar flexion assistance than younger subjects and that, apart from reducing the metabolic cost, the exoskeleton could also have an impact by improving gait efficiency in elderly (with reduced plantar flexion strength), similar to the positive effect in stroke patients [6].

We suggested that exoskeletons could improve quality of life, especially in elderly with reduced exercise tolerance, by improving walking speed, walking mobility, and physical activity [12–14]. In younger subjects, a reduction in metabolic cost with an assisting exoskeleton resulted in an increase in preferred walking speed [5] and our results suggest that this could be possible in elderly as well. However, it is possible that physical limitations, like plantar flexor weakness or stability, limit increased walking speed. Plantar flexor weakness is a determinant of the reduced step length and walking speed in elderly [18,19]. Previous attempts to improve plantar flexor strength to increase walking speed resulted in inconclusive findings [28] but as we showed that plantar flexion assistance with an exoskeleton improves step length, this emphasizes the role of the plantar flexors for speed generation and suggests that step length would not be a limitation to increase walking speed in healthy elderly. Dynamic stability is reduced in elderly [18] and variations in step length and step width are related to falls and mobility problems in the elderly [29]. Although stability was suggested to be a limiting factor in the application of exoskeletons [21], our population did not perceive the exoskeleton as disturbing their balance compared to walking with normal shoes and step width and intra-subject variability in step length and step width were not altered in the exoskeleton conditions.

One subject was not able to walk comfortably with the exoskeleton (Supplementary) due to knee flexion during stance, which increases metabolic cost [30] and perturbs exoskeleton assistance. This highlights the limitation of using one exoskeleton with general assistance. When populations with an increased degree of reduced exercise tolerance or reduced physical capabilities are studied, it will be important to gradually adapt the exoskeleton assistance to the needs of each subject and allow sufficient habituation. Differences in metabolic cost and perception between the first and second powered condition in our experiment suggested that habituation was not yet complete, at least in the first powered condition. Future research should therefore focus on individual optimization and habituation instead of using general population optimizations [1,23].

In conclusion, healthy and physically active elderly can walk with an ankle-foot exoskeleton that assists plantar flexion during the push-off. This causes a reduction in metabolic cost versus unpowered walking.

With improvements in design and actuation it should be possible to find reductions in metabolic cost versus walking with normal shoes. As we used physically active elderly, it is hard to generalize our results to elderly with an increased degree of reduced exercise tolerance or reduced physical capabilities. However, the next step is to study exoskeleton assistance in those populations as it is assumed that this could improve walking capacity. Nevertheless, our results in healthy and physically active elderly are promising for future applications and support to continue research of walking assisting exoskeletons in elderly and other populations with limited exercise tolerance.

Conflict of interest

None.

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GENERAL DISCUSSION

GENERAL DISCUSSION

1. Introduction

The overall goal of the thesis was to optimize the human-exoskeleton interaction to reduce the metabolic cost of walking so that exoskeletons can be used for practical applications. Therefore, we used a human experimental approach where metabolic responses to changes in different exoskeleton parameters were measured. By identifying required adaptation (*chapter 2*) and optimizing exoskeleton assistance parameters (*chapter* 3 and *chapter* 4), we used the exoskeleton to improve maximal (walking) performance (*chapter* 5) and to assist walking in healthy elderly (*chapter* 6).

Optimization of the human-exoskeleton interaction was possible by the development of a robust exoskeleton testbed (*chapter* 1). This exoskeleton testbed allowed to manipulate different exoskeleton parameters in a carefully controlled manner and allowed to measure the resultant human responses. By changing an assistive parameter over a certain range (e.g. the actuation onset timing), it was possible to detect the optimal value for that assistive parameter and improve our understanding of the relation between that specific exoskeleton parameter and the human response. Due to the stability of our exoskeleton testbed we were able to test a high number of conditions in one protocol (as far as we know more than in any other parameter sweep study at the moment) and a high number of subjects in the last years. The optimization of WALL-X from a technical point of view (*chapter* 1) is an important part of this thesis but will not be included in the discussion as these technical improvements were not a goal in itself but a sine qua non to perform the experimental studies.

In the first three studies, the exoskeleton parameters that were manipulated were chosen based on specific hypotheses to identify exoskeleton adaptation (*chapter* 2) and optimize exoskeleton actuation (*chapter* 3 and *chapter* 4). In the last two studies, the exoskeleton was used to answer more general research questions, using the exoskeleton to improve maximal performance (*chapter* 5) and as an application to assist elderly during walking (*chapter* 6).

2. Answers to research questions

This thesis aimed to answer specific research questions focused on adaptation, actuation, performance and applications. In this paragraph I aim to briefly answer these research questions using the information that was gathered in the different chapters.

2.1. Adaptation

> ? How long does metabolic adaptation to walking with an exoskeleton takes?

During continuously walking with WALL-X for 24 min, the metabolic cost gradually reduced until 18 ± 5 min, when a plateau was reached in the metabolic cost and subjects were considered metabolically adapted (*chapter* 2). The time to reach this adapted stated was referred to as the adaptation time. Although exoskeleton assistance resulted in a reduction in the metabolic cost compared to unpowered exoskeleton walking almost immediately, the reduction increased over the course of the adaptation until subjects were adapted and a metabolic reduction of 16% was found. The reported adaptation time of app. 20 min is much shorter compared to the reported adaptation times of more than 90 min for exoskeletons with a proportional EMG controller based on biological muscle activity. This suggests that it is easier to get used to walking with an exoskeleton with an on-off controller with an optimal actuation timing, where exoskeleton assistance is switched on during the push-off based on kinematic measures. These results could be used as adaptation guidelines for similar exoskeleton devices or future experimental work with WALL-X and other similar exoskeletons.

> ! Adaptation in the metabolic cost of walking with WALL-X takes app 20 min.

? Which neuromechanical changes occur in the leg muscles during this adaptation?

Within a few minutes, subjects walked kinematically similar to how they walk when adapted (*chapter* 2). However, while subjects seemed kinematically adapted after only a few minutes, EMG activity in the m. soleus, the m. tibialis anterior and the m. biceps femoris further reduced over the course of the adaptation. A neuromechanical adaptation process in the muscles occurred in which muscle activity reduced, leading to higher reductions in the metabolic cost after the adaptation time of app 20 min. This mechanism shows similarity with children which evolve from gross activation toward a more economical muscle activity production when they learn to walk (Okamoto et al., 2003) or the reduction in muscle activity while learning a task (Dugas and Marteniuk, 1989; Engelhorn, 1988).This shows how adaptation should not only be evaluated based on kinematics and exoskeleton kinetics and how neuromechanical adaptations on a muscular level seem related to the metabolic cost of exoskeleton walking.

! The adaptation in the metabolic cost is accompanied with a reduction in muscle activity.

2.2. Actuation

> ? Can actuation onset timing be optimized (independent of average positive exoskeleton mechanical power) in order to reduce the metabolic cost of exoskeleton walking?

An extensive parameter sweep experiment indicated that actuation onset timing was optimal when it started around 40% of the stride time for low, medium or high amounts of average exoskeleton ankle power (*chapter* 3). Earlier and later actuation timings resulted in an increased metabolic cost and the relationship between actuation timing and metabolic cost remained stable over different amounts of average exoskeleton power. The highest reduction in the metabolic cost was found with an actuation onset timing at 42% of the stride and medium exoskeleton power with a reduction in the metabolic cost for powered versus unpowered exoskeleton walking of 21%.

> ? Can average positive exoskeleton ankle mechanical power be optimized (independent of actuation onset timing) in order to reduce the metabolic cost of exoskeleton walking?

The parameter sweep also revealed that the metabolic cost of exoskeleton walking was lowest when average exoskeleton power was around 0.4 $W \cdot kg^{-1}$ for all actuation timings (*chapter* 3). Lower or higher amounts of average exoskeleton power resulted in an increased metabolic cost and this relationship between average exoskeleton power and the metabolic cost seemed stable across actuation timings. Higher amounts of exoskeleton power, at least with our device, did not result in higher reductions in the metabolic cost. The combination of the optimal actuation timing (starting around 40% of the stride time) and the optimal average exoskeleton power (around 0.4 $W \cdot kg^{-1}$) resulted in a reduction in the metabolic cost of 21% compared to unpowered walking and 12% compared to normal walking. It is believed that this suggested actuation timing and average exoskeleton power can also be used by other research groups to improve exoskeleton actuation, similar to the attention that was given to our previous findings on actuation timing (Malcolm et al., 2013). Our results indicate that the metabolic cost of exoskeleton walking depends both on the actuation onset timing and the average amount of exoskeleton power and that the metabolic cost of walking with exoskeletons can be reduced by optimizing both.

> ! Optimizing actuation onset timing and average exoskeleton power of WALL-X reduces the metabolic cost with more than 10% compared to normal walking.

?	What is the optimal actuation timing during uphill walking with an exoskeleton?

During uphill walking on a 15% inclination, an actuation timing starting around 30% of the stride is metabolically optimal with reductions of more than 12% versus unpowered walking (*chapter 4*). Earlier or later actuation onset timings resulted in slightly lower reductions in metabolic cost. These actuation timings could be used as actuation timing guidelines for uphill walking with WALL-X or other similar exoskeletons and showed that exoskeletons with plantar flexion assistance during the push-off can assist both level and uphill walking.

!	Optimizing actuation timing of WALL-X during uphill walking reduces the metabolic cost with more than 10 % compared to unpowered walking.

2.3. Performance

? Can an exoskeleton increase maximal walking performance?

During a maximal walking exercise test assistance of WALL-X reduced the metabolic cost, also when carrying heavy weights and when highly fatigued, with 8 to 10% compared to unpowered walking (*chapter* 5). Subjects were able to reach a maximal effort, which was evaluated based on physiological measures, both with a powered and with an unpowered exoskeleton. Due to the constant reduction in metabolic cost in the powered exoskeleton condition, subjects were able to continue the exercise test longer and increased maximal performance because of exoskeleton assistance with 8%. This study was the first to show how exoskeletons can increase maximal performance and reduce the metabolic cost of loaded walking, which increases the possible applications for assistive exoskeletons. These results are also relevant for military applications as soldiers often have to walk with loads up to 35 kg (Patterson et al., 2005).

! WALL-X improves maximal performance during uphill walking while carrying weight.

2.4. Applications

? Can an exoskeleton be used to reduce the metabolic cost of walking in healthy elderly?

In healthy elderly, the metabolic cost was reduced with 12% versus unpowered walking after only 20 min of adaptation but not when compared to walking with normal shoes due to the mild exoskeleton assistance that was chosen for this population (*chapter* 6). However, healthy elderly showed similar reductions compared to a healthy and young population, which confirms the applicability of assistive exoskeletons in elderly or other populations with reduced exercise tolerance to improve their walking capacity. Our results support to continue research and developing practical applications with exoskeletons for subjects with walking difficulties.

! WALL-X is a good tool to assist elderly during walking and reduces the metabolic cost.

3. The assistive mechanism

Apart from the specific research questions that were answered above, one of the research goals was to improve general understanding of the human-exoskeleton interaction. Recent improvements in exoskeleton design and control resulted in several exoskeletons that reduce the metabolic cost of exoskeleton walking below normal walking (Collins et al., 2015; Malcolm et al., 2013; Mooney et al., 2014a), which we also found with WALL-X (*chapter* 3). This difference between powered exoskeleton walking and normal walking is the result of the assistive effect of the exoskeleton and the metabolic penalty of walking with the unpowered exoskeleton. However, the exact mechanism of this assistive effect that causes a reduction in the metabolic cost between powered and unpowered exoskeleton walking is not (yet) entirely clear. The next paragraphs will mainly focus on that differences between powered and unpowered locomotion.

3.1. Metabolic cost of level walking

As exoskeletons reduce the metabolic cost of walking, it is likely that they influence the most important determinants that are responsible for the metabolic cost during walking. The contributions of the major determinants of the metabolic cost of walking differ between studies. Values between 37 and 65% are suggested for the step-to-step transition (Donelan et al., 2002a; Grabowski, A. et al., 2005; Kuo et al., 2005; Umberger, 2010), which is mainly determined by the centre-of-mass push-off work and the centre-of-mass collision work (Fig. 1). For leg swing, values between 10 and 30% are reported (Doke et al., 2005; Umberger, 2010) and around 30% or more for supporting body weight and controlling raising and falling of the bCOM during the single stance phase (Grabowski, A. et al., 2005; Neptune et al., 2004b; Umberger, 2010). There are also smaller contributions for active lateral stabilisation (Donelan et al., 2004) and swinging the arms (Collins et al., 2009; Meyns et al., 2013; Ortega et al., 2008). Some differences between studies could be attributed to recent findings that muscle work associated with the push-off also relies on isometric muscle contractions during the midstance phase, which also comes with a significant metabolic cost (Collins et al., 2015; Umberger, 2010)(Fig. 1). Also, recent evidence shows that an impulsive ankle push-off assists leg swing initiation (Lipfert et al., 2014), which indicates that a clear distinction between the different contributors of the metabolic cost of walking is difficult. Overall, a rough estimation indicates that more than one third of the metabolic cost of walking is linked to the step-to-step transition, a little less than one third is linked to leg swing cost, a little less than one third is linked to supporting body weight and the remainder is linked to arm swing, active stabilisation and other costs (Fig. 1). It is assumed that the same mechanisms determine the metabolic cost of unpowered exoskeleton walking as for normal walking, although the relative contribution of the different mechanisms can be slightly altered.

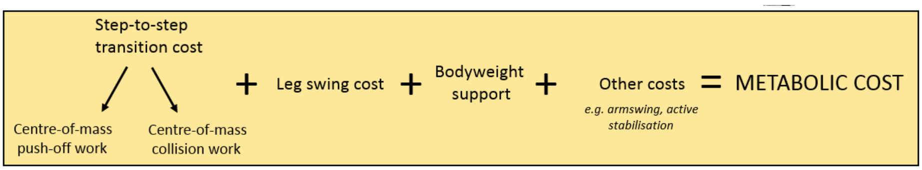


Fig. 1
Major determinants of the metabolic cost of walking.

3.2. Exoskeletons and the step-to-step transition cost

A first mechanism that can explain the reduction in metabolic cost for powered exoskeleton walking compared to unpowered exoskeleton walking is the influence of exoskeleton assistance on the centre-of-mass push-off work, which is mainly determined by the ankle joint positive work during the push-off (Kuo et al., 2005). When exoskeletons assist plantar flexion during the push-off, the exoskeleton replaces around half of the biological ankle work and thus reduces the contribution of the plantar flexor muscles (Gordon et al., 2006; Kao et al., 2010; Kinnaird and Ferris, 2009; Sawicki and Ferris, 2008, 2009c). Indeed, we found reductions in m. soleus and m. gastrocnemius muscle activity in the push-off during level walking with exoskeleton assistance (*chapter* 2 and *chapter* 3). As the plantar flexors represent around 20 to 25% of the metabolic energy during walking (Sawicki and Ferris, 2008, 2009c; Umberger and Rubenson, 2011), it is expected that ankle-foot exoskeletons can maximally reduce the metabolic cost with 20 to 25 % compared to unpowered walking if the exoskeleton replaces all the muscle work of the plantar flexors. This seems consistent with the reduction in the metabolic cost of 21% compared to unpowered exoskeleton walking (*chapter* 3) when the exoskeleton delivered around 100% of the normal biological positive ankle power during walking.

However, there are several indications that the assistive mechanism is more complex. First of all, the muscular activity of the ankle plantar flexors is reduced but still existing during exoskeleton assistance. The plantar flexors still have to perform considerable amounts of work. Second, we reported significant increases in the muscular activity of the tibialis anterior during exoskeleton walking (*chapter* 2 and 3), due to the increased plantar flexion in the push-off and the time to empty the pneumatic muscles. This increased activity of the m. tibialis anterior in the beginning of the swing phase should increase the metabolic cost of exoskeleton walking. Third, computer simulations revealed that part of the metabolic cost of the plantar flexors is consumed during the single stance phase, prior to the push-off (Umberger, 2010).

The latter was recently confirmed in an experimental exoskeleton study where Collins et al. showed how exoskeletons can also reduce the metabolic cost of walking by reducing the cost of isometric muscle contractions during the midstance phase (Collins et al., 2015). With a passive exoskeleton that uses a mechanical clutch to hold a spring during the stance phase and recoils it during the push-off (Fig 2A), a reduction in the metabolic cost of walking with 7% compared to walking with the exoskeleton without the elastic assistance of the spring was found (Fig 2C). The mechanism that caused the reduction was not attributed to directly assisting the push-off during the recoil of the spring but mainly to the torque assistance in the midstance phase by spring stretching, prior to the push-off. As the torque onset of WALL-X starts after midstance and does mainly assist concentric plantar flexion during the push-off, it seems unlikely that WALL-X also assisted the isometric muscle function during the midstance phase.

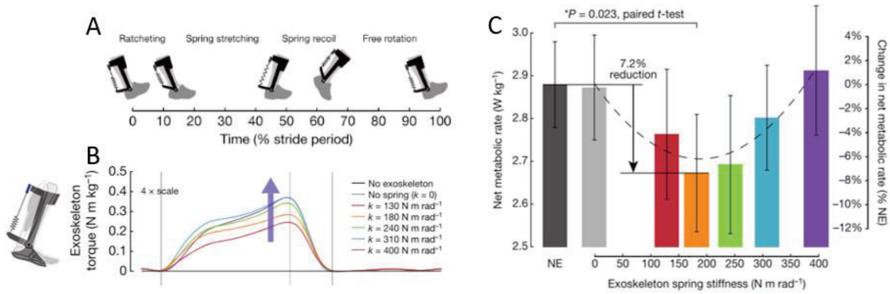
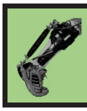


Fig. 2

Collins et al. developed a passive exoskeleton with a mechanical clutch mechanism and a spring (A). They studied walking with the exoskeleton with springs with differences in stiffness. The spring delivered torque during the stance phase (B), first by stretching the spring, and in the push-off by recoil of the spring. For an optimal stiffness this assistance resulted in a reduction in metabolic cost of 7% (C) compared to wearing the exoskeleton without the spring disengaged (0) and also a reduction of 7% compared to not wearing an exoskeleton (NE). The difference between wearing the exoskeleton with the spring disengaged and not wearing an exoskeleton was close to zero due to the low weight of the exoskeleton and the individually optimized fit. From Collins et al. (Collins et al., 2015).

All together, these findings indicate that exoskeletons that assist the plantar flexors during the push-off, like WALL-X, reduce the step-to-step transition cost for the user by reducing the biological ankle push-off work but that this mechanism is insufficient to explain the reduction in the metabolic cost of over 20% for powered exoskeleton walking compared to unpowered exoskeleton walking (*chapter* 3).

A second mechanism that can be related to reducing the step-to-step transition cost and thus the reduction in metabolic cost for walking with ankle-foot exoskeletons is a reduction of the centre-of-mass collision work. Based on the simplest walking model (Kuo, 2002), Malcolm et al. (Malcolm et al., 2013) found some indications that ankle-foot exoskeletons that increase the plantar flexion range of motion and velocity during the push-off, reduce the collision work by a more effective redirection of the centre of mass during the push-off. However, we did not find indications for reduced collision (*chapter* 3) and did not find reduced collision work in a recent pilot test (appendix 1). This is in accordance with prosthesis experiments (Caputo and Collins, 2014a) that did not find reduced collision with push-off assistance. Exoskeletons that assist plantar flexion during the push-off seem not to reduce the metabolic cost of walking by reducing the centre-of-mass collision work.



WALL-X reduces the effort of the plantar flexors during the push-off but this is insufficient to explain the reductions in the metabolic cost of 21% compared to unpowered walking.

3.3. Exoskeletons and the leg swing cost

The metabolic cost of walking is also resulting from the cost of swinging the legs (Doke et al., 2005; Gottschall and Kram, 2005; Griffin et al., 2003; Neptune et al., 2008; Umberger, 2010). Recently, it was suggested that the impulsive ankle push-off during walking also powers leg swing initiation (Lipfert et al., 2014). Indeed, we found reductions in m. rectus femoris and m. biceps femoris activity during exoskeleton assistance that are related to a more passive leg swing (*chapter* 2 and 3). A similar reduction in the muscular activity of the m. rectus femoris was found with an EMG-controlled exoskeleton (Koller et al., 2015) and also in a recent prosthesis study, powered ankle push-off reduced the leg swing cost (Caputo and Collins, 2014a).

In all our studies, exoskeleton assistance during the push-off increased the plantar flexion in the ankle joint and the plantar flexion velocity during the push-off. Several studies also reported an increase in total ankle joint power, which is the combination of the biological and the exoskeleton ankle joint power (Kao et al., 2010; Malcolm et al., 2013; Norris, Granata, et al., 2007) during exoskeleton walking. The more vigorous plantar flexion and the increased plantar flexion velocity seem key elements in the assistive mechanism of WALL-X that result in a reduction of the leg swing cost. The importance of the increased plantar flexion is emphasized by recent results of an autonomous ankle-foot exoskeleton that used a torque profile based on our previous results (Malcolm et al., 2013) but in combination with normal joint angles, leading to an increased metabolic cost instead of a reduction in the metabolic cost during exoskeleton walking (Van Dijk et al., 2014). This is important towards the development of future exoskeleton controllers. Pneumatic exoskeletons that assist plantar flexion during the push-off reduce the cost of swinging the legs by reducing the effort in the hip muscles and it seems like this requires an increased maximal plantar flexion and increased plantar flexion velocity during the push-off.

Exoskeleton assistance with WALL-X reduces the cost of leg swing through increased and fast plantar flexion.

3.4. Exoskeletons and other costs

There are also some other contributors of the metabolic cost of walking that could be influenced by exoskeleton assistance. As mentioned earlier, increased m. tibialis activity was found in all our studies and should lead to an increase in the metabolic cost. It is possible that the exoskeleton reduces the metabolic cost associated with body weight support because of a bracing effect around the ankle joint, but this is at the moment still unclear. Also, the effect on the cost for arm swing is rather unknown, while previous exoskeleton studies highlighted the possibility of reduced stability during exoskeleton walking (Norris, Marsh, et al., 2007), which will probably increase the metabolic cost associated with active lateral stabilisation (Donelan et al., 2004).

3.5. The assistive mechanism during level walking

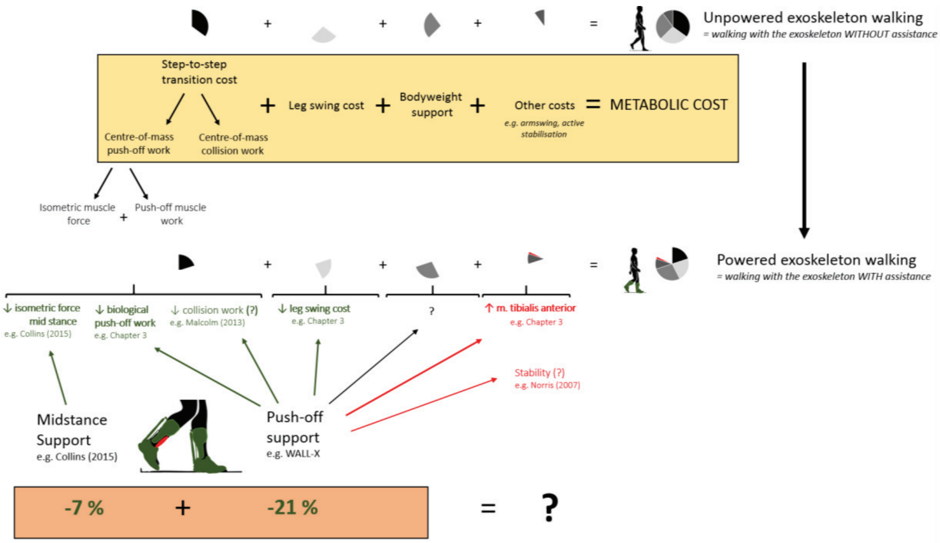


Fig. 3
Representation of the major determinants of the metabolic cost of powered and unpowered exoskeleton walking (respectively referring to exoskeleton walking with and without assistance) and how these determinants are influenced during exoskeleton assistance.

In conclusion, exoskeletons that deliver torque support during the midstance phase, can reduce metabolic cost of walking with 7% compared to unpowered walking (Collins et al., 2015). Exoskeletons that assist the plantar flexors during the push-off (like WALL-X) reduce the metabolic cost of walking by more than 20% by a combination of mechanisms. The step-to-step transition cost is reduced for the user by reducing the biological push-off work and possibly by reducing collision work. An important part of the reduction in the metabolic cost is attributed to reduced leg swing cost, for which increased plantar flexion and plantar flexion velocity seem necessary. While the influence on body weight support and arm swing are unclear, the plantar flexion assistance also leads to increased antagonistic m. tibialis anterior muscle activity and perhaps an increased metabolic cost due to reduced stability. Still, the assistive mechanism of WALL-X seems sufficient to reduce the metabolic cost with 21% compared to unpowered walking (*chapter* 3)(Fig. 3). The combination of early and midstance ankle torque support to unload the plantar flexion muscles during isometric contractions and powered ankle-push off assistance to reduce biological push-off work and leg swing cost could lead to even larger reductions in metabolic cost, especially as the passive torque support in the study of Collins et al. lead to inefficient, rapid shortening of plantar flexor muscles during the push-off (Collins et al., 2015). Powered assistance of the plantar flexors during this push-off phase could then compensate for this inefficiency.

There are also other possibilities to further reduce the metabolic cost of walking based on the suggested mechanisms of (powered) exoskeleton walking (Fig. 3). Focusing on the negative exoskeleton power that we saw in the beginning of the swing phase in most studies, could further reduce the metabolic cost of walking. This negative power seems disadvantageous and is linked with increased muscle activity in the m. tibialis anterior. Reducing this negative power with an earlier actuation ending or accelerated emptying of the pneumatic muscles might further reduce the metabolic cost of walking.

One might wonder if the reductions that we found are already close to the maximal reductions that are possible with ankle assistance. While we suggest that a combined approach of torque assistance during the midstance phase with powered plantar flexion, in combination with a lightweight exoskeleton design and reducing the negative exoskeleton power in early swing might further reduce the metabolic cost, it is unlikely that reductions of more than 30% are possible. While assisting other joints could further reduce the metabolic cost of walking, previous attempts to assist the knee or hip where not as successful as ankle assistance (Do Nascimento et al., 2008; Sawicki and Ferris, 2009b). The approach in soft exoskeleton suits, combining hip and ankle assistance (Asbeck et al., 2015) probably has the highest potential for further reducing the metabolic cost of walking. However, if reductions in the metabolic cost of 20% compared to normal walking could be realized, which seems realistic based on the reported results with WALL-X in combinations with recent developments in exoskeleton research, this seems sufficient to start to use exoskeletons as applications to assist humans in rehabilitation or in daily life.

Exoskeletons that combine torque support during single stance with powered assistance in the push-off might result in higher reductions in the metabolic cost of walking.

3.6. The assistive mechanism during uphill walking

Uphill walking is fundamentally different from level walking as mechanical work must be performed against gravity. Still, the ability to use plantar flexion assistance to reduce the effort at the hip that was found during level walking, was also found during uphill walking (*chapter* 4). In level walking, this was mainly related to reducing the leg swing cost. In uphill walking, it was suggested that ankle plantar flexion assistance was used to raise the body centre-of-mass and reduced the effort of the hip joint in the contralateral leg. While inverse dynamics seem necessary to confirm these findings, it appeared that plantar flexion assistance during uphill walking had minor effects on the muscular activity of the plantar flexors and it was suggested that the additional plantar flexion power of the exoskeleton was not only used to replace biological ankle joint work but also to increase total ankle joint work to raise the body centre of mass. This is probably done in an attempt to reduce the metabolic cost of uphill walking by reducing the effort of the knee- and hip extensors. This indicates that ankle plantar flexion assistance reduces the metabolic cost of walking not only by assisting the plantar flexor muscles but also by assisting more proximal muscles during uphill walking and emphasizes the ability of humans to adapt to mechanical assistance in an energetically beneficial manner.

Plantar flexion assistance with WALL-X during uphill walking is used to raise the body centre-of-mass in order to reduce the effort of the knee- and hip extensors.

3.7. Implications for exoskeleton research and applications

Future research should continue to focus on understanding the assistive mechanism of exoskeletons. While this could further increase the reductions in the metabolic cost, it is especially important if exoskeletons are to be used in impaired populations to make sure that assisting specific joints do not lead to compensation of the user in other joints, increased loads in other joints, or increased muscle activity in other muscles during walking with exoskeletons. Recently, Collins et al (Collins et al., 2015) and Farris et al. (Farris et al., 2013) highlighted the potentially negative effect of exoskeleton assistance on muscle fascicle lengths. This emphasizes the difficulty of optimal exoskeleton assistance due to the influence of elastic storage and return, bi-articular muscles and possible effects on muscle fascicle lengths. It seems hard to combine all these elements to optimize exoskeletons, but the systematic approach of varying certain parameters and measuring the human responses, which we did in our study and which is recently done in other exoskeleton studies (Collins et al., 2015; Ding et al., 2014) and prosthesis studies (Caputo and Collins, 2014a; Malcolm et al., 2015), seems a good way to improve our understanding of the human-exoskeleton interaction.

Exoskeletons are often seen as a tool to assist specific muscles in the rehabilitation, depending on the needs of the patient. We showed that the assistive mechanism of exoskeletons (Fig. 3) has an influence on several muscles and joints, which makes it a huge challenge to develop an exoskeleton that allows to assist specific muscles more isolated. This also had an influence on the research with exoskeleton at our lab. One of our initial exoskeleton goals was to use powered exoskeletons to study challenging topics in physiology as we assumed that an exoskeleton allowed to isolate the contribution of different muscles (by reducing or augmenting their effort) (Malcolm et al. 2009b) in the metabolic cost of different types of locomotion and allowed to determine the contribution of specific muscle groups to exhaustion and fatigue processes during human locomotion. However, the assistive mechanism seems to influence more than one joint and/or muscle groups, which makes this approach more difficult, which was one of the reasons why we started to focus on other research questions.

4. Beyond research questions

Apart from the specific research questions that were answered above, our research improves general understanding of the human-exoskeleton interaction and several conclusions could be made that were not included in our initial research questions.

4.1. Other aspects of adaptation

We studied adaptation (*chapter* 2) and concluded that adaptation time to WALL-X takes app. 20 min. These findings were confirmed in our other studies as we found fast and high reductions in metabolic cost after a prior adaptation time of app. 20 min during uphill walking (*chapter* 4), uphill walking with loads (*chapter* 5) and even in healthy elderly (*chapter* 6). A possible concern is to which degree adaptation needs to be condition-specific. It is unclear if adaptation to WALL-X is depending on the assistance pattern (e.g. actuation timing and average exoskeleton power) or the environment (e.g. walking speed, walking inclination, carried weight, etc.). We found fast adaptations to uphill walking (*chapter* 4) and uphill walking while carrying weights (*chapter* 5) after an adaptation to level walking. Also, we found fast adaptations to conditions with varying actuation parameters after an adaptation to one specific condition (*chapter* 4). Across all our experiments, subjects were able to easily walk with the exoskeleton after only a few minutes and even when randomized conditions were used with actuation settings going from a very early actuation to a late actuation and average exoskeleton power from low to high, subjects felt comfortable walking with the exoskeleton within several minutes. Together with my own experience with exoskeleton walking, this lead to the conclusion that adaptation to WALL-X is general, within certain limits. If one adapts to level walking, it will be easier to walk uphill with WALL-X, and if one adapts to WALL-X with specific actuation settings, it will be easier to walk with WALL-X with slightly altered actuation settings.

A second possible concern is to what degree adaptation needs to be repeated on an additional day. Due to a large number of experimental conditions (*chapter* 3) or limited endurance capacity, e.g. in elderly (*chapter* 6), it is not always possible to perform adaptation and experimental data collection on the same day. Previous research in unilateral exoskeletons indicated that adaptation was almost immediately effective on a second day after a prior habituation session (Cain et al., 2007; Gordon and Ferris, 2007; Kinnaird and Ferris, 2009). However, metabolic cost was not measured in these studies and findings in unilateral studies with an EMG-controlled exoskeleton cannot be used for bilateral exoskeleton studies with an on-off controller without caution. Therefore, we re-analysed the results of *chapter* 2 in which we let 10 subjects adapt to 12 powered exoskeleton conditions (3 min) on a prior day. On a second day we measured metabolic cost of the 12 powered conditions (4 min), that were applied in randomized order. Instead of averaging the metabolic cost of the powered conditions of this second day based on the actuation settings (e.g. averaging all the *EarlyLow* conditions, all the *EarlyMedium* conditions, etc.), we averaged the conditions according to the order in which they were applied (all powered conditions that were done first were averaged as the first trials, all the powered conditions that were done second were averaged as the second trials, etc.), independent of the actuation settings of the conditions. Due to the randomization one would suggest that the average metabolic cost of all the first trials would be the same as the average for all the second trials, all the third trials, etc. However, we found a higher metabolic cost in the first trials compared to the following trials (Fig. 4). Repeated measures ANOVA showed that the metabolic cost in the first trials was app. 15% higher compared to all following trials (except for trial 8 and 9), while no other trials where significantly different from each other ($P \leq 0.05$).

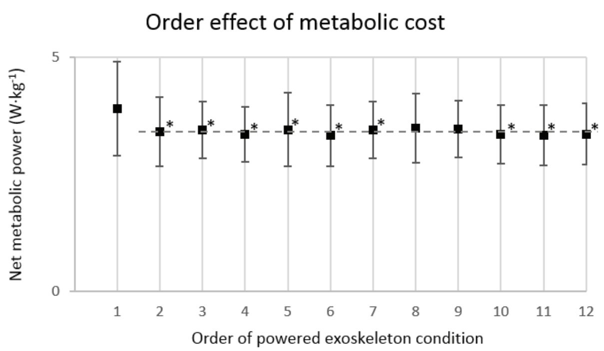


Fig. 4

The net metabolic power of the 12 powered conditions for 10 subjects (*chapter* 3). The metabolic cost is averaged for the order in which the conditions are done, independent of which powered conditions was done first. All powered conditions that were done first were averaged as the first trial, all powered conditions that were done second were averaged as the second trial, etc. * indicate a sign. difference between the average of the first trials.

This indicates that the metabolic cost in that first trial is on average increased as subjects need that first powered condition (independent from which condition it is) to adapt to exoskeleton walking, despite adaptation on a prior day. It is suggested that this did not strongly influence our results. Due to the randomisation and averaging over 10 subjects this 15% increase can influence the metabolic cost with around 1.5%. However, as also the unpowered condition can be higher, it is expected that this did not influence the differences between conditions.

A similar effect was seen in the elderly in *chapter* 6 as the reduction in the metabolic cost in the first powered condition (-9%) was on average lower compared to a second powered condition (-12%), which was also seen in the perception scores. Therefore, we conclude that subjects need about 4-5 min of adaptation on a second day to regain the adapted state after adaptation on a prior day.

A third aspect of adaptation is the effect of long-term use of exoskeletons, which we refer to as a training effect. Recently, exoskeletons became more robust and recent developments in exoskeleton research make it possible that exoskeletons will soon be implemented in a rehabilitation setting. It is likely that subjects then walk with an exoskeleton, several times a week for several weeks. The neuromuscular training effect to walking with an assistive exoskeleton over several weeks is unclear. It is possible that neuromechanical changes occur and further reduce the metabolic cost of exoskeleton walking, similar to the positive effect of power training on muscle force in the first weeks of a training period due to neural adaptations (Baechle and Earle, 2008). Although some of the few studies that focussed on exoskeleton assistance in stroke patients over several weeks reported improvements in walking speed and walking pattern (Agrawal et al., 2007; Fleerkotte et al., 2014; Krishnan et al., 2012), it is also possible that muscular and neuromechanical changes occur due to walking with the exoskeleton, that have a negative influence on normal walking without exoskeleton. It is therefore important that future research should focus on the training effect of walking with exoskeletons.

Adaptation time to WALL-X seems universal to a certain degree and a short adaptation (app. 5 min) seems necessary on a new day after a prior adaptation on a previous day.

4.2. Assistance pattern

A difficulty in exoskeleton research is the fact that carefully controlled experiments are time-consuming, especially when metabolic measurements are done as they require a condition duration of at least 4 min. It would be valuable to be able to predict the exoskeleton efficiency based on other parameters that can be determined before the start of an experiment. Sawicki and Ferris (Sawicki and Ferris, 2008) introduced the performance index to evaluate exoskeleton efficiency but this cannot be used to estimate the metabolic cost before the actual experiment. Mooney et al. (Mooney et al., 2014b) introduced the augmentation factor (AF), which estimates the change in metabolic power caused by an exoskeleton. The augmentation factor is calculated as follows:

$$AF = \frac{p^{+} + p^{dis}}{\eta} - \sum_{i=1}^{4} \beta_i m_i$$

The AF estimates the metabolic improvements of an exoskeleton in Watts, with a positive AF for a reduction in metabolic cost and a negative AF for an increase in metabolic cost. It takes into account the mean positive mechanical power supplied by the exoskeleton (p^{+}), the net mechanical power dissipation (p^{dis}), which is negative when the exoskeleton removes more mechanical energy from the body than it supplies, and the added mass on each limb segment (m_i) with the location of those masses (β)(Browning et al., 2007). Muscle-tendon efficiency (η) is used to convert mechanical power of the exoskeleton to metabolic power and is set at 0.41 based on literature data. This means that in previous exoskeleton studies adding 41W of exoskeleton mechanical power resulted in a reduction in the metabolic power of 100W. This augmentation factor appeared to be able to make a crude estimate of the metabolic improvements of exoskeleton walking for a broad range of powered and passive exoskeleton designs (Mooney et al., 2014b)(Fig. 5A). It suggests that exoskeletons should provide substantial positive mechanical power with minimal added leg mass and net negative power dissipation. As mentioned earlier, the weight of the device is an important factor in the metabolic penalty of wearing the exoskeleton and will have an influence on the metabolic cost as the metabolic reduction between powered exoskeleton walking and normal walking without exoskeleton will depend on the assistive effect of the exoskeleton and the penalty of wearing the exoskeleton due to the additional weight and possible movement restrictions.

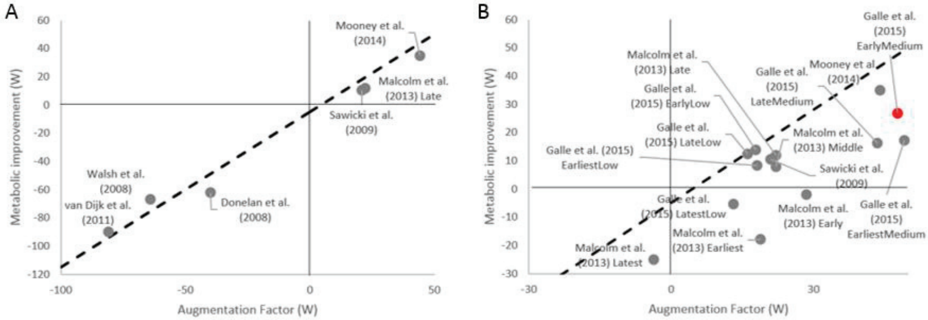
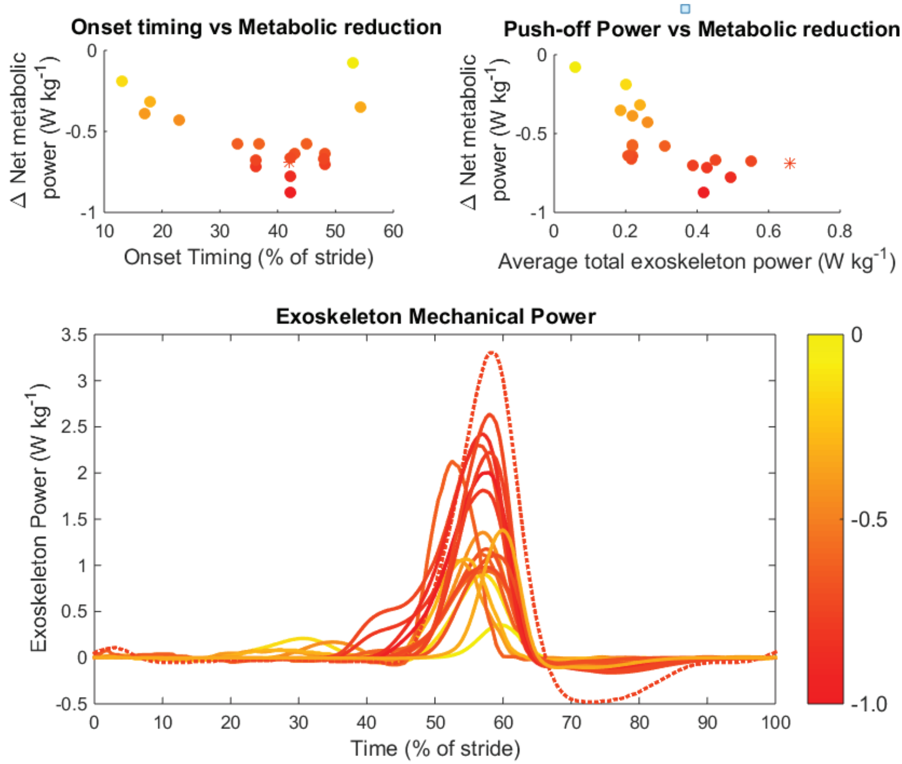


Fig. 5

Mooney et al (Mooney et al., 2014b) calculated the augmentation factor (A) for a passive exoskeleton that was designed to minimize joint work (van Dijk et al., 2011), a quasi-passive leg exoskeleton (Walsh et al., 2007), an energy-recycling knee harvester (Donelan et al., 2008) and the powered ankle-foot exoskeletons of Sawicki et al. (Sawicki and Ferris, 2009c) and Malcolm et al. (Malcolm et al., 2013). We added the results of Malcolm et al. (Malcolm et al. 2013) and the results of *chapter* 3 for all powered exoskeleton conditions (B).

However, the reduction in the metabolic cost for exoskeletons during walking will not only depend on the weight of the device, but also on the assistance of the exoskeleton. The AF includes the positive exoskeleton power and the power dissipation but does not limit the amount of power or include actuation timing guidelines, which we showed to be important determinants of the metabolic cost of walking (*chapter* 3). While this AF is valuable to evaluate a broad range of devices (Fig. 5A), their estimations of reported results are sometimes rough and when our results with powered exoskeletons are added to the predictions, it shows that the AF is not able to estimate the metabolic cost for the tested changes in actuation parameters of *chapter* 3 (Fig. 5B).



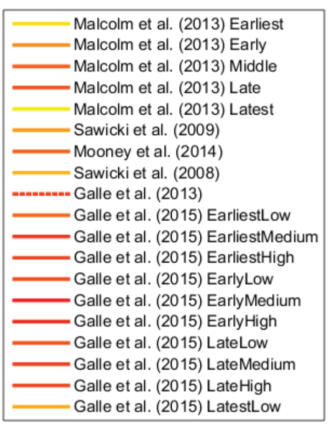


Fig. 6

Combined figure of the result of all the powered exoskeleton conditions and the resulting reduction in metabolic cost for powered versus unpowered walking of *chapter* 2 and 3, the results of Malcolm et al. (Malcolm et al., 2013), Sawicki et al. (Sawicki and Ferris, 2008, 2009c) and Mooney et al. (Mooney et al., 2014a). The different conditions of Malcolm et al. (Malcolm et al., 2013) are called Earliest, Early, Middle, Late and Latest according to the actuation timing onset. Several studies that examined the effect of walking with powered exoskeletons on metabolic cost of walking were combined to establish the relationship between actuation onset timing (A), average positive exoskeleton ankle power (B) and the exoskeleton mechanical power curve (C) on the reduction in metabolic cost for exoskeleton walking. Actuation onset timing (A) was determined based on the published exoskeleton power curves, (B) average positive exoskeleton ankle power for a stride was determined based on the published exoskeleton power curves (B) and summed for both legs. The exoskeleton power curves (C) were digitized based on the published figures and the reduction in metabolic cost for powered versus unpowered locomotion was calculated based on averages if not explicitly stated in the articles. Colours of the dots and the power curves are depending on the reduction in metabolic cost of powered versus unpowered walking. We only focused on the metabolic cost of powered versus unpowered locomotion as exoskeleton assistance will influence the differences in metabolic cost between powered and unpowered locomotion, which is referred to as Δ net metabolic power. It is assumed that the exoskeleton power of Galle et al. (Galle et al., 2013) is overestimated due to a damaged load cell (see *chapter* 1 for more details) and this study is therefore marked with a *.

The importance of actuation onset timing and the average exoskeleton positive ankle power is emphasized when other exoskeleton studies are added to the results of *chapter* 3 (Fig. 6). The second order relationship for actuation timing and the reduction in the metabolic cost for powered versus unpowered locomotion is still valid when results of other studies are added. Also the exponential relationship between average positive exoskeleton ankle power (combined for both legs in accordance with *chapter* 2) and the reduction in metabolic cost for powered versus unpowered locomotion is still valid when other exoskeleton studies are added, indicating that the average amount of power is limited for reductions in the metabolic cost. These findings indicate that a good estimation of the reduction in metabolic cost for exoskeleton walking compared to normal walking should include the weight of the device, but also specific assistance parameters from which actuation timing and average positive exoskeleton power seem important. With the pneumatic muscles that are used in WALL-X, it is hard to carefully control other assistance parameters. However, it is likely that other parameters, e.g. concerning the shape of the exoskeleton power curve or relating to the maximal ankle plantar flexion during the push-off (which we suggested to be a key aspect for the reduction in leg swing cost earlier in this discussion), do also influence the metabolic cost of exoskeleton walking. The high number of potential parameters that influence the metabolic cost of exoskeleton walking makes it a huge challenge to determine the metabolic cost of walking based on exoskeleton assistance parameters or exoskeleton weight. At the moment, experimentally optimization of exoskeleton actuation by varying certain parameters and measuring the human responses seems still necessary to evaluate the effect of exoskeleton assistance on the metabolic cost of walking.

At the moment, experimental studies are still necessary to establish the effect of exoskeleton assistance parameters on the metabolic cost of exoskeleton walking.

4.3. Exoskeleton applications in patient populations

Exoskeletons could be an assistive tool for subjects with walking difficulties and have the advantage that they promote active participation of the user and should on the long term allow to walk on stairs, in narrow passages, and on rough terrain. They could become a natural assistance in the rehabilitation and in daily life and could deliver assistance as needed. Several applications are suggested and experimentally tested in the last years.

Exoskeletons are often suggested to be a tool for gait assistance in stroke survivors during rehabilitation (Takahashi et al., 2015) or to restore normal gait in patients with spinal cord injury (Duerinck et al., 2012; Swinnen et al., 2010). Several studies showed that whole-body interventions with exoskeletons have positive effects on clinical tests, walking speed and the walking pattern during a rehabilitation over several weeks in stroke survivors (Agrawal et al., 2007; Krishnan et al., 2012) and in patients with spinal cord injury (Fleerkotte et al., 2014). However, the added value compared to standard rehabilitation is sometimes questioned (Swinnen et al., 2010). Also, the feasibility of smaller ankle-foot exoskeletons as a tool to enhance paretic ankle moment and gait symmetry is shown in stroke patients (Rossi et al., 2015; Takahashi et al., 2015) and as a tool to improve ankle push-off in patients with incomplete spinal cord injury (Sawicki et al., 2006). These smaller exoskeleton have the benefit that walking remains more natural and that, with the recent development of several autonomous exoskeletons (Asbeck et al., 2015; Collins et al., 2015; Mooney et al., 2014a), they can be used outdoors and in daily life. Also, they can be especially useful in patients who already regained limited mobility because they have less kinematical restrictions and additional mass compared to whole-body exoskeletons (Rossi et al., 2015). Future research should focus on the effect of long-term interventions with ankle-foot exoskeletons and the long-term effect on the unassisted walking pattern.

We showed the feasibility of ankle-foot exoskeletons in healthy elderly, older than 65 (*chapter* 6). Our results in healthy elderly support to continue research in elderly as they could potentially benefit from exoskeleton assistance to improve physical activity if walking becomes easier or if they can walk faster with the same effort, similar to the positive effects of a power assisted bicycle on physical activity in elderly (Louis et al., 2012). While these result are promising, caution is necessary on balance and potential falls, especially in elderly or patients populations. It is shown that walking requires active lateral stabilisation (Donelan et al., 2004) but walking impairments are often accompanied with increased lateral pelvis movements (De Bujanda et al., 2003) and reduced stability (Bruijn, Sjoerd M. et al., 2013), which can predict future falls (Hilliard et al., 2008). Trunk movements have an influence on step width (Hurt et al., 2010) and are related with balance deficits (Adkin et al., 2005). Therefore, future research with exoskeletons in impaired populations should also include stability measures (Bruijn, S M et al., 2013) and hip and trunk movements in order to evaluate balance during walking.

In the introduction we suggested that ankle-foot exoskeletons could also be useful in other patients with reduced exercise tolerance in general, or patients with Chronic Obstructive Pulmonary Disease (COPD) in specific. These patients could benefit from a reduction in the metabolic cost or an increase in walking speed or walking duration because the room for improvement by traditional rehabilitation therapy is limited. Our promising results in healthy elderly suggest that COPD patients could benefit from exoskeleton assistance although COPD patients have some specific physiological characteristics and can only walk for short durations, which can influence the human-exoskeleton interaction. Therefore we performed a pilot test in 3 COPD patients to study the feasibility of the use of exoskeletons in COPD patients (appendix 2). We showed that COPD patients that follow a post-rehabilitation program can reduce the metabolic cost of powered exoskeleton walking below the cost of unpowered exoskeleton walking. Metabolic cost was not reduced below normal walking due to the high penalty of wearing the unpowered exoskeleton and the relatively low reductions in the metabolic cost for powered versus unpowered walking compared to our results in healthy elderly in *chapter* 6.

It should be noted that the metabolic measurements of these COPD patients showed some differences from those in a healthy population, which might be due to their disease or lower physical function. As most COPD patients are unable to continuously walking for long durations, it is questionable if metabolic measurements are the best measures to evaluate efficiency of exoskeletons. Another measure could be the O_2 saturation, which decreased below 90% for the 2 most severe COPD patients during normal walking.

Still, we can conclude that this pilot test shows the feasibility of the use of ankle-foot exoskeletons in COPD patients and supports to continue research with exoskeletons in this population. A first step should be to reduce the penalty of wearing the unpowered exoskeleton by reducing the weight and improving the design of the exoskeleton. We used one exoskeleton for all our patients but recent exoskeleton studies make custom exoskeleton frames for every subject (Collins et al., 2015; Takahashi et al., 2015). A good compromise would be to use one exoskeleton with a lower weight that allows more individual adjustments. While individual exoskeletons are useful for experimental research, it is probably too expensive to do this in a realistic rehabilitation setting. A second step should be to increase the reduction in metabolic cost between powered and unpowered locomotion. This could be done by improving the assistance parameters, as suggested earlier, or by individual optimization of the assistance as it is possible that these populations show more individual differences. Another important aspect would be to increase adaptation time as it seems evident that these patients needs longer adaptation to exoskeleton walking compared to a young and healthy populations due to the reduced motor control that comes with age (Ketcham and Stelmach, 2004).

If the metabolic cost of exoskeleton walking can be further reduced for COPD patients, exoskeletons could be used during the rehabilitation and increase the assisted walking speed, the walking duration, and the amount of uninterrupted walking bouts with less breathlessness. This can have a positive impact on motivation and it is possible that this has a positive impact on rehabilitation outcomes. Additionally, if we can demonstrate how COPD patients can benefit from exoskeleton assistance, this opens perspectives towards the development and the implementation of autonomous exoskeletons for daily life assistance in patients with reduced exercise tolerance. This could further improve mobility, participation in society and hence quality of life.

COPD patients can walk with WALL-X. Future research should focus on further reducing the metabolic cost by optimizing the exoskeleton, the assistance and the adaptation.

5. Future directions

In this thesis, we mainly focussed on improving the understanding of the human-exoskeleton interaction. The carefully controlled experiments that we did are difficult and time-consuming. Small mechanical differences can cause substantial changes and it is hard to predict how humans will respond to mechanical assistance of exoskeletons. A first approach for future research is to continue research in which exoskeleton parameters are varied, based on specific hypotheses to influence one of the main determinants of the metabolic cost of exoskeleton walking (Fig. 3). In these studies the human response to changes in these exoskeleton parameters should be measured in order to improve the understanding of the human-exoskeleton interaction as there is still much to be known to advance exoskeleton research. This can help to improve the predictions of human responses and further reduce the metabolic cost of exoskeleton walking by further optimizing exoskeleton assistance.

A second approach for future exoskeleton research is to further reduce the penalty of wearing the unpowered exoskeleton. Weight is an essential factor in exoskeleton research and the added weight should always be minimized as additional weight, especially when distally located, largely increases the metabolic energy expenditure and can influence walking biomechanics (Browning et al., 2007). Even when focussed on the comparison between powered and unpowered conditions, the aim should be to remain limb weight as close as possible to walking without the exoskeleton. Recent developments in exoskeleton designs showed how the difference in the metabolic cost between unpowered exoskeleton walking and normal walking could be below 5%, even for fully autonomous devices (Collins et al., 2015; Mooney et al., 2014a), while the penalty of wearing WALL-X is often around 10%. This implies that reductions of more than 15% for powered exoskeleton walking compared to normal walking, are possible if the penalty of walking with WALL-X could be further reduced.

The increase in the metabolic cost for wearing WALL-X unpowered is over 10% while current exoskeletons can reduce this penalty below 5%.

A third approach is to continue improving exoskeleton hardware and control in order to make exoskeletons autonomous. Recent developments in exoskeleton research resulted in several fully autonomous ankle exoskeletons, referring to the fact that all actuators and power supply are carried by the user, which increases the possible applications of exoskeletons in general as outdoor use and applications in daily life become realistic (Fig. 7).

Fig. 7
Example of recent autonomous ankle exoskeletons from left to right: the powered autonomous exoskeleton of van Dijk et al. (Van Dijk et al., 2014), the soft autonomous exoskeleton suit of Asbeck et al. (Asbeck et al., 2015), the powered autonomous exoskeleton of Mooney et al. (Mooney et al., 2014a) and the passive autonomous exoskeleton of Collins et al. (Collins et al., 2015).

van Dijk et al. (Van Dijk et al., 2014; van Dijk et al., 2014) built a fully autonomous ankle-foot exoskeleton (Fig. 7) that aimed to deliver similar torques compared to WALL-X (Malcolm et al., 2013) but they did not find a reduction in the metabolic cost versus a zero-work condition (where the exoskeleton was worn but did not deliver any power), probably because they did not impose fast plantar flexion during the push-off. Most current exoskeletons use rigid links (Collins et al., 2015; Malcolm et al., 2013; Sawicki and Ferris, 2008; Takahashi et al., 2015), which induces difficulties to align the exoskeleton joint with the biological joint and difficulties due to the inertia (Asbeck et al., 2014). Soft-exoskeleton suits (Fig. 7) can overcome these problems as they are light, rely on the biological joints to transmit torques, can be worn under normal clothes and have little or no effect on normal walking kinematics (Asbeck et al., 2014). Such an autonomous soft-exoskeleton suit can assist the ankle and hip joint during walking and reduces the metabolic cost with 6 to 10% compared to walking with the exoskeleton without actuation but not yet compared to normal walking (Asbeck et al., 2014, 2015; Wehner et al., 2013). Mooney et al. (Mooney et al., 2014a, 2014b) developed the first autonomous ankle-foot exoskeleton that reduces the metabolic cost of walking (Fig. 7). It consists of a pair of fiberglass struts, an actuator mounted on the anterior shank and a battery and control package worn on the waist. This new design uses the biological ankle joint as the rotation joint for the exoskeleton. If the actuation profile of WALL-X could be implemented in this exoskeleton, it is suggested that reductions in the metabolic cost of more than 15% compared to normal walking should be possible. Another successful approach is to use passive devices, which reduces the weight as there is no need for a heavy power source. A passive exoskeleton was previously shown effective in reducing muscle activity and energy cost in hopping (Farris et al., 2013; Ferris, Bohra, et al., 2006; Grabowski, A. M. and Herr, 2009). Collins et al. developed a fully autonomous passive ankle exoskeleton (Fig. 7) that stores energy in a spring during the stance phase and re-uses it during the push-off. They used our adaptation time (*chapter* 2) and found reduction in the metabolic cost of 7% compared to unpowered exoskeleton walking but due to the low weight of the device, they also found a reduction of 7% for exoskeleton walking versus normal walking without the exoskeleton as the penalty of wearing the exoskeleton was negligible. These results emphasize that elastic elements can play an important role in future exoskeletons. When combined with powered assistance they can reduce power requirements and allow to place actuators on metabolically beneficial locations (Asbeck et al., 2015). Malcolm et al. (Malcolm et al., 2013) even suggested to recycle energy from contralateral knee swing with a clutch mechanism to power the push-off and a similar energy-recycling approach was used to generate electricity during walking by harvesting energy during periods of negative power in the knee, although this slightly increased metabolic cost (Donelan et al., 2008; Li et al., 2009).

Recent improvements in exoskeleton design and control resulted in fully autonomous exoskeletons that reduce the metabolic cost of normal walking.

Despite the introduction of several autonomous exoskeletons, an exoskeleton testbed like WALL-X with off-board hardware will remain useful in future exoskeleton research to study the human response to changes in exoskeleton assistance. While the pneumatic muscles of WALL-X limit the number of parameters that can be controlled, Jackson et al. (Jackson and Collins, 2014) developed an exoskeleton testbed that uses Bowden cables, which improves control of exoskeleton assistance parameters. Several labs use an exoskeleton or prosthesis testbed (Caputo and Collins, 2014a; Ding et al., 2014; Malcolm et al., 2015) to improve the understanding of the human-exoskeleton interaction and further optimize exoskeleton assistance before actually building lightweight and autonomous devices. The human-exoskeleton interaction remains extremely complex and one of the best ways to improve the understanding is by measuring the human response to changes in several exoskeleton parameters. Therefore, it is useful to first develop and use an exoskeleton testbed that allows to manipulate several exoskeleton parameters over a broad range and optimize exoskeleton assistance before autonomous devices are built. The best example is the prosthesis that was developed by Collins and Kuo (Collins and Kuo, 2010) based on an energy recycling approach, in which energy that is normally dissipated as negative work in the collision could be recycled and used as additional positive work during the push off. A lot of effort was invested in this prototype that successfully reduced the metabolic energy cost in subjects with an artificially impaired ankle (Collins and Kuo, 2010). Unfortunately, this was not found in real amputees (Segal et al., 2012) and lead the development of a prosthesis testbed (Caputo and Collins, 2014b) that allows to test certain hypotheses before actually building usable prostheses.

WALL-X and similar tethered exoskeleton testbeds will remain important in future exoskeleton research to further optimize exoskeleton assistance and exoskeleton design.

A fourth aspect for future research is to improve the evaluation of exoskeleton concepts. We proposed to continue performing parameter sweeps but the number of conditions that could be tested is limited and still, this approach is slow and time-consuming. One way to overcome this time-consuming approach is the use of simulation studies. Although experimental studies will remain necessary to confirm findings of simulations studies, they could help to explore specific parameters in order to choose experimental conditions. A good simulation model of exoskeleton walking would increase the speed of exoskeleton research with a huge amount. Van Dijk et al. (van Dijk and van der Kooij, 2013)(Fig. 8) simulated exoskeleton walking based on the neuromuscular model of Geyer and Herr (Geyer and Herr, 2010) and added an exoskeleton torque based on a precious exoskeleton study (Cain et al., 2007). Based on the model of Umberger (Umberger et al., 2003), the metabolic cost was estimated. While the resulting model was able to find stable gait patterns and the muscle activation patterns showed some similarities with experimental data, the energy expenditure estimations did not strongly coincide with experimental findings. Afschrift et al. (Afschrift et al., 2014) showed how simulations could help to estimate assistance as needed for muscle weakness with exoskeletons. With the huge amount of experimental data that is published in the last years, also in our lab, it seems possible to develop improved simulation models that allow better predictions of the metabolic cost of exoskeleton walking.

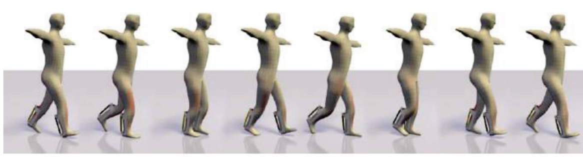


Fig. 8
Example of the simulation of exoskeleton walking. From van Dijk et al. (van Dijk and van der Kooij, 2013).

Another approach the improve the speed in exoskeleton research is to use an approach where instantaneous energetic cost is estimated (Selinger and Donelan, 2014), perhaps combined with a body-in-the-loop approach (Felt et al., 2014) where certain parameters could be optimized and responses could be evaluated much faster. In *chapter* 3 and *chapter* 6 we also measured perception, which could also be a possible measure to evaluate exoskeleton assistance faster compared to metabolic measurements. These visual analog scales are used in a wide variety of research areas to evaluate pain (Bijur et al., 2001; DeLoach et al., 1998), anxiety (Kindler et al., 2000), shortness of breath (Ander et al.,

2004) or comfort in footwear (Mills et al., 2010). However, caution is necessary to interpret these values (Torrance et al., 2001) and although some studies showed the reliability and validity of these measures for breathlessness and general fatigue during a submaximal exercise (Grant et al., 1999) it would be useful to evaluate the reliability and validity of these measures in an exoskeleton context in the future. A third way to speed up the exoskeleton evaluation process might be to use summed EMG values instead of respiratory measurements to estimate the metabolic cost (Jackson and Collins, 2015; Marsh and Martin, 1997; Silder et al., 2012).

A last focus of future research would be to increase the use of exoskeleton in a rehabilitation setting or in (simulated) daily life applications for different populations, to establish the true potential of exoskeletons and long-term effects of exoskeleton assistance. After all, more evidence is necessary on the added value of exoskeletons in rehabilitation and in daily life and the possible influence on quality of life for different patient populations. If exoskeletons are shown effective in improving quality of life for specific populations, exoskeleton research will receive more attention.

Future exoskeleton research should focus on demonstrating the effect of exoskeletons on quality of life for specific patient populations

6. Limitations of the research

When overlooking the different studies that were done, several limitations could be formulated, from which we will discuss a few.

Most of the data analysis only focusses on sagittal plane kinetics and kinematics. However, walking is not a one directional movement and hip and trunk movements in the frontal and transverse plane are also important, especially as hip and trunk movements play a role in balance control (Winter, 1995). Some of our findings could be the result of trunk and hip movements in these planes.

Concerning our experimental protocol we often did not include a standard shoe condition to limit the number of experimental conditions. Although we estimated the differences between exoskeleton walking and walking in standard shoes based on previous results and literature, the inclusion of a standard shoe condition would have clarified the relevance of our results.

The unreliable load cell data in some of our studies limited the understanding of the assistive mechanism and the use of an instrumented treadmill would also have improved the understanding of the assistive mechanism of exoskeletons.

We used a limited number of subjects in all studies that were homogenous in age, weight and length. While this was done to be able to use one exoskeleton for all subjects, it is possible that lower reductions in the metabolic cost would be found if a representative sample of the entire populations was used. We also excluded subjects where data collection was insufficient due to technical failure of the exoskeleton. If exoskeletons will be used for practical applications, it is necessary that they become more robust.

We used an exoskeleton with off-board hardware. While this was useful to test a high number of experimental conditions, it is not sure if the same results could be found for autonomous exoskeletons. While we assume that recent lightweight and autonomous exoskeletons should be able to deliver the same assistance compared to WALL-X, experimental validation seems necessary to confirm these hypothesis. The use of our pneumatic muscles also limited to control of the assistance parameters and they have limited possibilities for autonomous exoskeletons.

While several other limitations could be formulated, we think that the limitations of our studies are acceptable, given the experimental nature of this research. We are confident that our findings are reliable and we were able to confirm several findings in more than one study.

7. General conclusion

In general, this thesis aimed to improve the understanding of the human-exoskeleton interaction in order to optimize exoskeleton assistance for applications in healthy and impaired subjects. It seems fair to conclude that this thesis improved the understanding of the human-exoskeleton interaction, resulting in significant reductions in the metabolic cost of walking and the demonstration of the feasibility of exoskeletons for applications in healthy subjects to improve maximal performance, to reduce the metabolic cost in elderly and a first step towards demonstrating the feasibility in subjects with reduced exercise tolerance. We hope that our findings can be used by others to improve exoskeletons so that one day subjects can benefit from exoskeleton assistance during ambulant walking. However, several questions remain unanswered and future exoskeleton research is necessary to meet these targets.

In this thesis, an exoskeleton testbed was used to manipulate exoskeleton assistance parameters. By focussing on the metabolic response to changes in the exoskeleton actuation parameters, we highlighted several important assistance parameters. We showed that adaptation to exoskeleton walking takes app. 20 min, which was used by other research groups to develop fully autonomous and passive exoskeletons that reduce the metabolic cost of walking. We showed how actuation timing and exoskeleton power could be optimized, leading to reductions of more than 21% versus unpowered exoskeleton walking and more than 12% versus normal walking. These actuation guidelines can now be implemented in other exoskeletons to further reduce the metabolic cost of walking. By optimizing the exoskeleton assistance we showed how exoskeletons could be used to increase human performance, which increases the possible applications of exoskeleton devices, and how exoskeleton could be used to assist elderly and other populations with reduced exercise tolerance to reduce the metabolic cost of walking. Future research should built on these results and further optimize exoskeleton actuation and use exoskeletons in populations with reduced walking related exercise tolerance to regain walking capacities and improve quality of life.

Somewhere in the near future, exoskeletons will allow us to "*work it harder, make it better, do it faster, makes us stronger*" and will improve the quality of life of certain populations and we hope that this thesis can contribute to these goals.

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PUBLICATIONS

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A1 ARTICLES – PUBLISHED

Galle, S., Malcolm, P., Derave, W., De Clercq, D. (2015) Uphill walking with a simple exoskeleton: Plantarflexion assistance leads to proximal adaptations. *Gait Posture*: 41 (1): 246-251.

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Galle, S., Derave, W., Bossuyt, F., Calders, P., Malcolm, P., De Clercq, D. (2015). Walking with a plantar flexion assisting exoskeleton in healthy elderly. *Gait Posture* (under review).

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[Oral presentations] - International

Galle, S., Malcolm, P., De Clercq, D. (2015) Walking with a plantar flexion assisting ankle-foot exoskeleton in an older population. *In proceedings of the 25th Conference of the International Society of Biomechanics (ISB)*; Glasgow (Scotland), Jul 12-16.

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[Poster presentations] - International

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[Hardware demonstrations] - International

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APPENDIX 1

BODY CENTRE-OF-MASS WORK

To improve the understanding of the assistive mechanism of ankle-foot exoskeletons and in preparation of future experiments, we performed a pilot test with 1 subject walking on an instrumented treadmill with WALL-X at 1.25 $m \cdot s^{-1}$. We calculated body center-of-mass power and body center-of-mass work for the collision phase, the rebound phase, the preload phase and the push-off phase (Fig. 1).

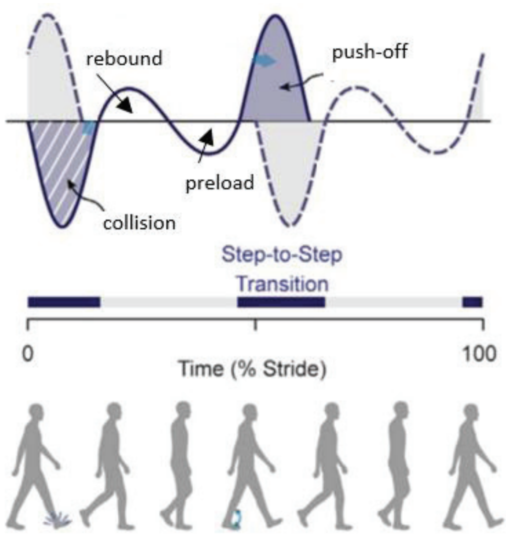

Fig. 1
Representation of the body center-of-mass power curve during a stride of the left (dark blue) and right leg (dashed blue) with the work bursts during the collision, rebound, preload and push-off phase. Adapted from Collins & Kuo (Collins and Kuo, 2010).

The subject walked during conditions of 2 min: once with the exoskeleton unpowered and three times with the exoskeleton powered. In the powered conditions the exoskeleton assisted plantar flexion with an actuation timing starting at 40% of the stride and with compressed air supply set at 2, 3 and 4 bar. These air pressure values were estimated to result in an average amount of exoskeleton positive power of 0.2, 0.3 and 0.4 $W \cdot kg^{-1}$ (summed for both legs in accordance with *chapter* 3). We calculated body center-of-mass power based on the ground reaction forces of each leg separately over the duration of a stride using the individual limb method (Donelan et al., 2002b). Body center-of-mass work was then calculated as the area under the positive and negative bursts of the power curve for the collision phase, the rebound phase, the preload phase and the push-off phase and averaged for both legs (Donelan et al., 2002a, 2002b)(Fig. 2). The results of the pilot test indicate that collision work is not reduced during exoskeleton assistance. Rebound work seems reduced during exoskeleton assistance and preload and push-off work seem increased.

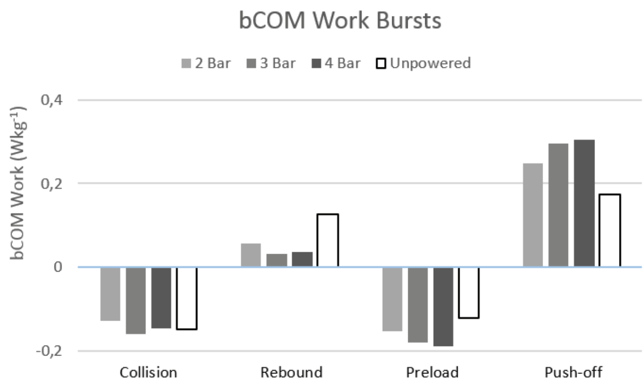

Fig. 2

Results of a pilot test: body center-of-mass (bCOM) work bursts based on the bCOM power curve for 1 subject walking at 1.25 $m \cdot s^{-1}$, once with an unpowered exoskeleton and three times with a powered exoskeleton where air pressure in the pneumatic muscles was set at 2, 3 and 4 bar. Positive and negative work bursts of the individual leg were calculated and averaged for the collision phase, the rebound phase, the preload phase and the push-off phase.

APPENDIX 2

EXOSKELETON ASSISTANCE IN COPD PATIENTS

We performed a pilot test in 3 COPD patients to study the feasibility of the use of exoskeletons in COPD patients (Fig. 3).

Fig. 3
Subject with COPD walking with WALL-X during a short walking conditions in our pilot test. The subjects were asked to give a sign with their hand when breathing became difficult and if they needed a break. In between walking conditions the face mask that was used to measure metabolic energy cost was removed in order the allow normal breathing and recover from the walking condition.

We tested 3 COPD patients (weight: 71.2 ± 4.9 kg; length: 171.0 ± 1.0 cm; European shoe size 43.3 ± 2.1, age 60 ± 5.3 y) that were following a post-rehabilitation program to retain their rehabilitation improvements at the University Hospital in Ghent. Two subjects were in good physical health and were able to walk during 5 min at 4 $km \cdot h^{-1}$ on the treadmill. For one subject this was divided into two 2.5 min sessions as this subject was unable to continuously walk for more than 3.5 min due to his disease. All subjects did a habituation to normal walking (5 min) on treadmill, unpowered walking on treadmill (5 min) and powered exoskeleton walking (5 min) on a first day. On a second day subjects performed 3 powered exoskeleton conditions (5 min each). On a third day, subjects performed the experimental protocol and O_2 consumption and CO_2 production were measured with a gas analyser. The experimental protocol consisted of 4 conditions of 5 min (or 3.5 min for one subjects): walking with normal shoes, with the unpowered exoskeleton, and twice with the powered exoskeleton (in semi-randomized order). In two subjects we found a reduction in the metabolic cost between powered and unpowered walking of 3% in the first powered condition and of 7% in the second powered condition. The penalty for wearing the exoskeleton was 15%, which explains why we did not find a reduction versus normal walking but a penalty of 7% versus normal walking instead. The third subjects was only able to walk for 3.5 minutes before shortness of breath arose. Metabolic cost was calculated based on the last 30s values and indicates a reduction of 4 and 7% versus unpowered walking and a penalty of 3 and 0% versus normal walking.